FAKULTETA ZA ELEKTROTEHNIKO
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Imre Cikajlo

SENZORNA INFORMACIJA PRI PONOVNEM UČENJU HOJE

DOKTORSKA DISERTACIJA

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Slike

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Povzetek

Hoja je ena najpomembnejših človekovih dejavnosti. V grobem jo razdelimo v dve fazi, fazo opore in fazo zamaha. Vsako od omenjenih faz lahko razdelimo še na podfaze. Tako je faza zamaha, ki bo predmet obravnave disertacije, razdeljena v dvig prstov, začetni zamah, osrednji zamah, končni zamah in dotik s peto. Taka delitev je primerna za analizo hoje. Če gre za gibalno nemoteno osebo, dobimo pri hoji lepo izražene posamezne faze. Ljudem, ki so doživeli poškodbo hrbtenjače oziroma osrednjega živčnega sistema, je hoja popolnoma ali delno odvzeta in so večinoma uporabniki invalidskih vozičkov. V primeru da lahko hodijo z uporabo ortoz, bergel ali hodulje, posamezne faze niso več izrazite, nasprotno, največkrat so popačene. Osebe z nepopolno poškodbo hrbtenjače hodijo s spremenjenim vzorcem hoje, saj jim je poškodba odvzela možnost upravljanja posameznih mišičnih skupin. Uporaba funkcionalne električne stimulacije omogoči obnovitev nekaterih motoričnih funkcij. Druga težava, ki spremlja ljudi s poškodbo hrbtenjače, je izguba senzorne informacije. Osebe nimajo povratne informacije o stanju svojih okončin, razen če jih vizualno spremljajo, kar pa povzroči neprimerno držo med hojo in bistveno zmanjšano hitrost hoje.

Pričujoča disertacija obravnava hojo oseb z nepopolno poškodbo hrbtenjače. Namen študije je razviti rehabilitacijski pripomoček in metodo, ki bi omogočala izboljšanje kvalitete hoje in morebiti skrajšala čas potreben za rehabilitacijo. Študija hoje oseb brez gibalnih motenj nudi vpogled v parametre, ki so podlaga za razvoj senzornega sistema. Senzorni sistem bi nadomestil prizadet naravni senzorni kanal in bi nudil potrebno senzorno informacijo za kvantitativno vrednotenje hoje. Ovrednotena hoja lahko služi za kasnejšo analizo in ugotavljanje določenih deficitov, svoj pravi namen pa doseže, če jo uporabimo za krmiljenje ali povratno informacijo. Razvoj originalnih algoritmov za detekcijo faze zamaha predstavlja osnovo, na kateri gradimo algoritem za vrednotenje zamaha. Na podlagi kvantitativnega vrednotenja je določena kognitivna povratna zanka, ki po opravljenem koraku obvesti osebo o ustreznosti zamaha, da lahko le-ta ustrezno ukrepa pri izvedbi naslednjega. Tak pristop omogoči zavestno sodelovanje

pacientov v rehabilitacijskem procesu. Večina pacientov uporablja funkcionalno električno stimulacijo kot pripomoček pri restavraciji motoričnih funkcij spodnjih ekstremitet. Vodenje stimulacije je lahko prepuščeno pacientu, fizioterapevtom ali računalniku. Ker se pri rehabilitaciji oseb z nepopolno poškodbo hrbtenjače na tekočem traku pojavi potreba po ponovljivosti, pride do izraza vodenje na podlagi senzorne informacije. Tako načrtovan sistem za ponovno učenje hoje vodi do končnega cilja, tj nadzorovano izboljšanje vzorca hoje, ki se kaže v zmanjšani potrebi po motorični podpori s funkcionalno električno stimulacijo.

Delo je razdeljeno v tri dele. Analiza hoje pri osebah brez gibalnih motenj nam služi kot referenca za razvoj modelov, na podlagi katerih določimo posamezne faze hoje in načrtamo algoritme za detekcijo in vrednotenje faze zamaha. Uporaba nevronskih mrež, Kalmanovega filtra in ostalih orodij za procesiranje signalov omogoča tvorjenje uporabljene senzorne informacije. Drugi del opisuje načrtovanje kognitivne povratne zanke in funkcionalne električne stimulacije kot pripomočka za obnovitev motoričnih funkcij pri paretičnih osebah. V tretjem delu je predstavljen sistem za ponovno učenje hoje pri paretičnih osebah.

Originalni prispevki disertacije so:

- razvoj algoritma za detekcijo faze zamaha iz posameznih senzorjev, pospeškometrov in žiroskopa
- razvoj algoritma za oceno kvalitete zamaha, ki predstavlja kompleksno informacijo iz preprostih senzorjev, žiroskopa, pospeškometrov in goniometra
- razvoj algoritma za določanje relativnega položaja okončine v zamahu na podlagi informacije iz preprostih senzorjev, žiroskopa in pospeškometrov,
- razvoj originalnega pristopa pri rehabilitaciji pacientov z nepopolno poškodbo hrbtenjače ali hemiplegijo z uporabo FES sistema, ki omogoča realizacijo gibov paralizirane ekstremitete preko perifernega vzbujanja spinalnega živčnega sistema, in senzornega sistema, ki nudi kognitivno povratno informacijo o stanju spodnje okončine oz kvaliteti zamaha pri hoji in lahko tudi aktivno sodeluje pri vodenju električne stimulacije.

Abstract

Walking is one of the most important human activities. In general it consists of two phases, a swing and stance phase. Both phases can be divided into sub-phases. The swing phase, which is the main topic of dissertation, is further divided into toe off, initial swing, middle swing, terminal swing and heel contact. The above classification is suitable for gait analysis. These phases are quite clear in a case of unimpaired person, while in impaired person, with spinal cord injury or injury of the central nervous system, they cannot be easily distinguished. Persons with complete spinal cord injury are wheelchair users, while persons with incomplete spinal cord injury may have preserved their mobility and often use crutches and various orthoses. In incomplete spinal cord injury the gait pattern is changed due to the ability to maintain control over several muscle groups. The use of functional electrical stimulation enables the restoration of some motor functions. Another problem those persons are dealing with, is a loss of sensory information. They have no feedback information about their lower extremities except visual control that is a cause of inappropriate poise during gait and walking speed decrement.

The objective of the dissertation is the analysis of the gait of the incomplete spinal cord injured persons. The purpose of the study is to develop a new rehabilitation aid and a method that will significantly improve the gait quality and might shorten the rehabilitation process. The outcome of the unimpaired human gait analysis are parameters useful for the development of a sensory system. The sensory system is intended for impaired natural sensory substitution and could be used for quantitative gait evaluation. The gait evaluation can be an applied in an off-line gait analysis and determination of gait deficits and can be used in control or cognitive feedback. The development of an algorithm for swing phase estimation is based on original algorithms for swing phase detection. On the basis of quantitative evaluation we could determine the cognitive feedback and therefore make the person aware of the swing adequacy when the swing is finished, so that the person is able to take prompt action. The presented approach enables to include the voluntary activity of the patients into rehabilitation process. Most of the patients use the functional electrical stimulation to restore motor functions of the lower extremities. The stimulation control can be managed by the patient himself or either let to the physiotherapist or to the computer. Since there is a need for repeatability in the rehabilitation of incomplete spinal cord injured persons, the advantages of the sensory supported control are introduced. The purpose of the presented gait re-education system is a gait pattern improvement that results in the decrease of necessary functional electrical stimulation.

The thesis is divided into three main sections. The unimpaired persons gait analysis is a reference for gait kinematics and dynamics modelling that present the basis of swing phase detection algorithms and swing phase estimation algorithm development. The use of neural networks, Kalman filter and other signal processing tools make possible the determination of the sensory information. The second section describes the cognitive feedback design and functional electrical stimulation as an orthotic device for the restoration of motor functions in incomplete spinal cord injured persons.

The main contributions of the thesis are as follows:

- the development of the swing phase detection algorithm for each sensor group, gyroscope and accelerometers ,
- the development of the original swing phase estimation algorithm that is based on assessed sensory information and forms the complex sensory information ,
- the development of the algorithm that estimates the relative position of the lower extremity,
- the development of the original approach in incomplete spinal cord injured persons rehabilitation using the FES system that enables the realization of the paralyzed extremity movement by exciting the peripheral spinal nerve system and multisensor system providing cognitive feedback information on swing phase quality during walking and is also an active part of the electrical stimulation control.

1

Uvod

Doktorska disertacija obravnava senzorno informacijo pri hoji. V uvodnih poglavjih so predstavljeni funkcionalni gibi, v primeru ko pride do poškodbe hrbtenjače in njihova obnovitev. V nadaljevanju sledi povzetek metod, s katerimi je možno nadomestiti izgubljeno naravno senzorno informacijo. S podrobnejšo analizo hoje z različnimi tipi senzorjev določimo potrebni nabor senzorjev. Originalni algoritmi za detekcijo faze zamaha in ocenjevanje kvalitete zamaha predstavljajo pomemben korak pri ocenjevanju hoje. Tak algoritem je bil implementiran v sistem za ponovno učenje hoje. V sistemu za ponovno učenje hoje nastopa tudi kognitivna povratna zanka, ki nudi uporabniku, pacientu in fizioterapevtom, informacijo o zamahu v obliki zvočnega signala. Ker je sistem primarno namenjen uporabi pri rehabilitaciji oseb z nepopolno poškodbo hrbtenjače, je uporaba funkcionalne električne stimulacije kot motoričnega pripomočka smiselna. Vrednotenje posameznih algoritmov, kakor tudi celotnega sistema, je rezultat številnih meritev na Inštitutu za rehabilitacijo v Ljubljani.

1.1 Funkcionalni gibi pred in po poškodbi hrbtenjače

Človeški organizem je zasnovan v smislu kompleksno zgrajenih sistemov, ki so med seboj povezani v celoto. Ogrodje organizma tvori mišičnoskeletni sistem, kakor že ime pove, sestavljen iz kosti, ki so medsebojno povezane s sklepi, ligamenti in mišičnokitnimi skupinami. Nad mišičnoskeletnim sistemom bedi živčni sistem, razvejan v množice perifernih živčnih regulatorjev, sestavljenih iz receptorjev, mreže nevronov in mišic. Delovanje takih regulatorjev opišemo s predstavitvijo osrednjega živčnega sistema kot procesne enote, ki z raznimi receptorji zajema senzorne informacije in jih vodi po aferentnih živčnih vlaknih. Ustrezne reakcije na sprejete dražljaje pošlje po eferentnih živčnih vlaknih do mišic in žlez. V mišicah in žlezah se nahajajo receptorji, ki sklenejo regulacijsko zanko. Delovanje take regulacije lahko ovira bolezen ali poškodba, kar se odraža kot nesposobnost človeka za izvajanje koordiniranih gibov ekstremitet. Take ekstremitete so deloma ali v celoti hrome, oseba pa paralizirana [1]. Posledica poškodbe hrbtenjače je prekinitev nevronske povezave med možgani in mišicami, ki izvršujejo gibe. Višina poškodbe ali lezije hrbtenjače določa tudi stopnjo poškodbe in določa, katere mišične skupine so prizadete. Gornje ekstremitete oživčujejo živci iz vratnega predela hrbtenjače, spodnje ektremitete pa živci, iz nižjega prsnega ali ledvenega predela hrbtenjače. Popolna poškodba v vratnem predelu ima za posledico tetraplegijo, kar v praktičnem pomeni omejen nadzor nad zgornjimi in nobenega nadzora nad spodnjimi ekstremitetami. Nižja poškodba hrbtenjače v prsnem ali ledvenem predelu ima za posledico paraplegijo, kjer je ohranjen nadzor nad zgornjimi ekstremitetami in trupom, izgubljen pa nadzor nad spodnjimi ekstremitetami. Večina tetraplegikov in paraplegikov s poškodbo gornjega motoričnega nevrona imajo ohranjen refleksni lok, ki se zaključuje v hrbtenjači. Spodnji motorični nevroni so sicer odrezani od možganov, ki tako nimajo več nadzora nad njimi, a kljub temu ohranijo skupaj z mišicami svojo funkcionalnost. Nevšečnosti se pojavljajo zaradi neaktivnosti paraliziranih ekstremitet [2]. V relativno kratkem času se pojavijo atrofija in kontrakture mišic v teh ektremitetah ter osifikacija sklepov in demineralizacija kosti. Te nevšečnosti spremljajo potem še dodatne notranje težave kot so funkcionalnost urinarnega trakta, slaba prekrvavitev paraliziranih ekstremitet, prebavnega trakta in ostalih notranjih organov. Obstaja pa tudi možnost preležanin [1, 2]. Funkcionalna električna stimulacija (FES) ponuja možnost obnovitve nekaterih gibalnih funkcij pri osebah s poškodbo hrbtenjače [3]. Funkcionalni gib ekstremitete dosežemo z električnim dražljajem, s katerim vzdražimo živec, s tem pa povzročimo kontrakcijo mišice oz mišične skupine [4]. Glede na način stimulacije ločimo dva različna tipa stimulacije, eferentni tip, kjer stimuliramo živec, ki vodi do mišice in aferentni tip, pri katerem stimuliramo periferni živec za prenos informacije v spinalni center, to pa vzbudi signal v motoričnem živčnem vlaknu, kar povzroči pokrčenje mišičnih vlaken in posledično funkcionalni gib ektremitete. Za razliko od funkcionalnega giba lahko FES uporabimo tudi v terapevtske namene. Izboljša prekrvavitev mišic in gibljivost sklepov. Slaba stran FES je predvsem utrujanje, ki je veliko hitrejše in drugačno, kakor pri naravnem gibu ekstremitet [5]. Zmanjšanje utrujanja je možno s spreminjanjem stimulacijskih parametrov ali z uporabo izmenične stimulacije posameznih mišičnih skupin [6].

1.2 Senzorji pri hoji

Kadar imamo v mislih vrednotenje človeške hoje, najprej pomislimo na merjenje vidnih parametrov hoje. Tukaj nastopajo spremembe položaja, hitrosti in pospeškov posameznih segmentov ekstremitet. V ozadju pa se dogajanjo še drugi procesi, ki jih človeško telo zaznava na različne načine in z različnimi senzorji. Hojo omogoča dejavnost mišic, aktivno krčenje in pasivno raztezanje, posledično pa se odraža to še v spremembi krvnega tlaka, hitrosti dihanja, na koži in drugih fizioloških dejavnikih. Glede na raznolik zajem signalov iz različnih virov lahko razdelimo senzorni sistem v tri kategorije [7]: eksteroceptivni, proprioceptivni in interoceptivni.

Eksteroceptivni senzorji zajemajo zunanje dražljaje, to so vid, sluh, senzorji za dotik in kemične dražljaje. Proprioceptivni nudijo informacijo o relativnem položaju dela ekstremitete glede na ostale segmente in o položaju telesa v prostoru. Interoceptivni senzorji pa spremljajo krvni tlak, koncentracijo glukoze in ostale spremembe znotraj telesa. Za analizo gibanja oz. hoje lahko zajemamo signale iz naravnih senzorjev [8], ki jih poseduje človeško telo ali za svoje potrebe namestimo umetne senzorje za zajem potrebnih veličin.

1.2.1 Naravni in umetni senzorji

Na področju biomedicinske tehnike je pogosta uporaba raznih vrst elektrod, s katerimi zajemamo biosignale. Tako vrsto senzorjev lahko imenujemo naravni senzorji. Uporaba naravnih senzorjev zna biti precej zapletena, saj je potrebno informacijo, ki jo nudijo, zajeti in pretvoriti v ustrezen električni signal, ki ga lahko ovrednotimo. Bioelektrični potenciali so posledica elektrokemijskih aktivnosti posameznih vrst celic, ki jih poznamo pod imenom vzdražljive celice, ki sestavljajo živčno, mišično tkivo. Električno gledano dajejo mirovni potencial, ko so vzdražene pa akcijski potencial. Meritve na aferentnih perifernih živcih [9, 10, 11, 8, 12] (elektronevrografija ENG) dajejo napetosti nekaj μV [13]. Frekvenčni razpon meritev je med 1 in 10 kHz, medtem ko je maksimalna moč signala pri 3 kHz. ENG signali vsebujejo nizke frekvence, ki jih je težko meriti zaradi prisotnosti električnih potencialov (elektromiogram EMG) sosednjih mišic. Največja težava pri meritvah živčnih signalov je nizko razmerje signal/šum (S/N). V μV območju signalov je lahko celo šum ojačevalnika večji od termičnega šuma živčnih elektrod, ki znaša 0.7 μV RMS [13].

Elektrode, s katerimi merimo biopotenciale znotraj telesa, so perkutane in interne elektrode. ENG elektrode so interne elektrode, kakor imenujemo elektrode, ki so kirurško nameščene pod kožo. Obravnavane elektrode, ki merijo signale neposredno na živcih, imajo posebno obliko tulca (cuff). Ovite so okrog debla merjenega živca, na katerem se senzorske aktivnosti odražajo v obliki električnih potencialov [13]. Veliko bolj enostavno je merjenje s površinskimi elektrodami, ki jih pritrdimo na kožo nad mišico. Najpogosteje uporabljena površinska elektroda je metalna elektroda. Da se izognemo potencialnemu pragu med kožo in elektrodo uporabljamo elektrolitski gel. Take vrste elektrod se uporabljajo za merjenje EMG (elektromiogram) in EEG (elektroencefalogram). Pri merjenju EMG so za elektrode v uporabi predvsem nerjaveče jeklo, platina ali pozlačeni materiali, predvsem kot preventiva proti kemijski reakciji z gelom, ki lahko povzroči kožne reakcije.

Za razliko od naravnih senzorjev imamo pri umetnih že zagotovljen električni izhodni signal, ki ga lahko vzorčimo, filtriramo, ojačimo in digitalno predelamo v želeno obliko. Pri hoji merimo predvsem kinematične veličine, kot so translacijska in kotna hitrost, pospešek, kot, nagib in položaj. Pri tem uporabimo celo paleto senzorjev, od pospeškometrov, žiroskopov, inklinometrov, goniometrov, do uporovnih elementov, optičnih merilnikov razdalje, položaja, Hallovih celic[14, 15, 16, 17, 18, 19, 20, 21, 22, 23, 24]. Kadar želimo še dinamično analizo hoje zajemamo še signale pritiskovnih plošč, sile in navore, ali merilnikov sile, ki jih namestimo na podplat. Take vrste senzorji zahtevajo za zanesljive meritve predvsem dobro namestitev na telo osebe, katere hojo želimo meriti. Zaradi njihove končne velikosti in predvsem neustrezne oblike, je namestitev zahtevna. Pri tem nas ovira tudi koža, ki onemogoča natančno zunanjo namestitev. Zato se včasih za namene regulacije ortoz poslužujejo minimizacije senzorjev in namestitve na skelet osebe [25, 26]. To pa zahteva drag operacijski postopek. Alternativa majhnim senzorjem pri meritvah so optični merilni sistemi (Optotrak®, Vicon®, Elite®), vendar je uporaba zaradi visoke cene omejena na laboratorijsko delo oz posebne kineziološke centre. Uporaba cenenih senzorjev zahteva dobro dodelane algoritme, ki dodobra odpravijo napake zaradi nenatančne namestitve.

1.2.2 Od merjene veličine do uporabnega podatka

Signali, ki jih zajamemo s senzorji, so navadno prevedeni v števila, ki še ne predstavljajo nobene uporabne informacije, zato jih je potrebno obdelati. Obdelava merjenih veličin pripelje do podatkov, uporabnih za določanje želenih informacij, s katerimi postrežemo uporabniku ali na podlagi katerih vodimo določen proces.

Če se omejimo na področje hoje in obdelave zajetih signalov, pridemo do zaključka, da večina dosedanjih algoritmov temelji na določanju faz hoje [27] in lahko služijo tudi za zaprtozančno vodenje električne stimulacije. Večina sistemov za električno stimulacijo je uporabljala nožno ali ročno prožilno tipko za proženje električne stimulacije [28, 2, 29, 30, 31]. Tong [32] je predlagal uporabo nevronskih mrež, ki bi 'nadomestile' ročno proženje. S trinivojsko strukturo mreže in večjo skupino senzorjev je izdelal ustrezni krmilnik hoje. Kostov je [33] uporabil in dodelal metode strojnega učenja z uporabo nevronskih mrež za vodenje funkcionalne električne stimulacije pri hoji. Informacije o hoji je zajemal s pritiskovnimi senzorji, nameščenimi v vložku čevlja, in goniometri v sklepih prizadete ektremitete. Uporabljeni metodi strojnega učenja sta služili za vodenje električne stimulacije v trenutku, ko je pacient hoteno sprožil stimulacijo s pritiskom na tipko. Podoben pristop je bil uporabljen pri korekciji padanja stopala. Klasični FES sistem [34] s petnim stikalom in enokanalnim peronealnim električnim stimulatorjem je nadomestil merilnik ENG (elektronevrogram) na suralnem živcu in implantiran peronealni stimulator. Vodenje stimulatorja temelji na procesiranju signalov. V ta namen so bili uporabljeni razni algoritmi z učenjem logičnih nevronskih mrež ali adaptivnih omejitvenih pravil. Kostov [35] je pokazal prednosti in slabosti enih in drugih metod. Z izbiro referenčnega signala je naučil nevronsko mrežo, da je iz signala med hojo razbrala fazo zamaha in fazo opore. Za preciznejše vodenje stimulacije, predvsem ko gre za večkanalno električno stimulacijo, razdelimo fazo zamaha in opore na podfaze. Pappas in skupina iz ETHZ, Švica, [36] so z uporabo pritiskovnih senzorjev, nameščenih v podplat obutve, in žiroskopom izdelali zanesljiv sistem za zaznavanje posameznih faz hoje. Algoritem z omejitvenimi pravili je omogočal razpoznavanje faze zamaha, faze opore, odriv in dostop. Za določanje časovnih dogodkov pri hoji je možno uporabiti tudi teorijo mehkih množic ali mehke logike [37]. Tak pristop je predvsem računsko manj zahteven, kar je prikladno za aplikacije v realnem času. Nadzorovano strojno učenje z odločitvenimi pravili pri določanju faz je pri hoji s pospeškometri opisal Williamson [38]. Willemsen [22] je s pomočjo dveh parov pospeškometrov določil pospešek v gleženjskem sklepu, slednjega pa uporabil za krmiljenje električne stimulacije. Precej preprostejši, a kljub temu učinkovit, algoritem je predstavil Dai [16]. Temeljil je na merilnikih nagiba, ki jih je namestil na segmente okončine, katerih nagib je meril. Stimulacijo je sprožil določen nastavljiv prag nagiba segmenta ekstremitete. Izklopila se je, ko je nagib padel pod drug nastavljiv prag. V času trajanja stimulacije je sistem ignoriral ostale zajete informacije, ki bi morebiti lahko sprožile stimulacijo. Možnost ponovnega proženja je bila omogočena šele po končani stimulaciji. Veltink [24] je uporabil enoosne pospeškometre za opisovanje statičnih in dinamičnih aktivnosti pri vsakodnevnih opravilih, kot so stoja, hoja, hoja po stopnicah, kolesarjenje. Meritve so se istočasno shranjevale v prenosnem sistemu, kasneje pa prenesle v računalnik, kjer je potekala analiza izmerjenih rezultatov. Kalibracijo pospeškometrov pri merjenju hoje je predstavil Lötters [19], medtem ko je Tong [21] pri analizi hoje uporabil žiroskop.

Večina opisanih principov temelji na uporabi preprostih cenenih senzorjev in kompleksnejših algoritmov. Tak pristop danes omogoča poceni zmogljiva računalniška oprema, medtem ko ostajajo precizni senzorji še vedno zelo dragi. Na tak način lahko dosežemo učinkovit merilni sistem, saj je cilj vsakega merilnega sistema zadostiti zahtevani točnosti in preciznosti.

1.3 Obnovitev vzorca hoje po poškodbi hrbtenjače

Vse metode merjenja in analize gibanja so bile primarno namenjene restavraciji hoje po poškodbi oz bolezni. Razvoj analize hoje in funkcionalne električne stimulacije sta bila vedno neposredno povezana. Brez analize gibanja tudi ni bilo moč načrtovati vzorcev, s katerimi bi stimulirali živce, z namenom doseči želen funkcionalen gib. Za analizo gibanja obstajajo kineziološki centri, opremljeni z optično merilno opremo (Optotrak® Northern Digital Inc., FastTrack®, Vicon®) in pripadajočo programsko opremo za analizo kinematičnih parametrov. Optično merilno opremo dopolnimo še s pritiskovnimi ploščami (AMTI®) in tako dobimo še podporne sile in momente. Na ta način lahko določimo posamezne faze hoje, ki jih potrebujemo pri obnovitvi hoje in zgradimo kinematični in dinamični model [39, 1]. Cenovno ugodnejšo in manj zahtevno analizo hoje zadovoljijo opisani ceneni senzorji in pripadajoči algoritmi, ki privedejo do uporabne informacije [40]. Z zadostnim naborom pridobljenih informacij lahko hojo analiziramo do nivoja, ki nam omogoča tudi vodenje funkcionalne električne stimulacije [41].

Funkcionalna električna stimulacija je zelo učinkovit pristop pri restavraciji hoje po poškodbi hrbtenjače ali možganski kapi. Z električno stimulacijo aktiviramo motorično funkcijo ekstremitet, ki je odvzeta ali motena zaradi poškodbe ali bolezni. Kljub temu obstajajo razlike v strategiji in pristopu pri obnovitvi stoje, vstajanja in hoje plegičnih (para, tetra) in paretičnih oseb. Vstajanje plegičnih oseb s pomočjo rok [1, 42] se razlikuje po številu uporabljenih stimulacijskih kanalov, stimulacijskem vzorcu, ki ga lahko določimo po predhodni meritvi trajektorije težišča telesa, in fizični pripravljenosti osebe. Obnovljena hoja s funkcionalno električno stimulacijo se razlikuje od naravne hoje po načinu krmiljenja, povratne informacije, pač glede na ohranjene sposobnosti osebe, katere hojo obnavljamo [43]. Zato obnovitev hoje v grobem razdelimo glede na poškodbo hrbtenjače. Pri plegičnih osebah je motorična funkcija popolnoma odvzeta, medtem ko je pri paretičnih osebah delno ohranjena, odvisno od višine lezije.

1.3.1 Obnovitev vzorca hoje pri plegičnih osebah

O paraplegiji, govorimo takrat, ko je poškodba v prsnem ali ledvenem predelu hrbtenjače popolna. To pomeni, da ni nobene povezave med spodnjim delom hrbtenjače in možgani in oseba nima nobenega nadzora nad spodnjimi ekstremitetami. Z uporabo FESa obnovimo motorične funkcije potrebnih mišičnih skupin [28]. Predpogoj je seveda zadostna ojačitev atrofiranih mišic, ki jo zagotovimo s terapevtsko električno stimulacijo takoj po poškodbi.

Druga težava, ki se pojavi pri popolni poškodbi hrbtenjače, je povezana s senzornim sistemom. S popolno poškodbo je odrezan tudi senzorični del živčnega sistema, tako da oseba tudi ničesar ne čuti, torej nima nobene povratne informacije, razen vizualne. Vizualna povratna informacija lahko služi kot povratna informacija pri odprtozančnem krmiljenju FESa. Predprogramirano sekvenco stimulacijskih pulzov sproži pacient kar s pritiskom na prožilno tipko. Zaprtozančna regulacija zahteva namestitev umetnih senzorjev ali merjenje signalov iz senzoričnih živcev. Tak pristop na podlagi izmerjenih parametrov hoje proži FES. Seveda je najteže izbrati strategijo, ki uporablja zadostno ali optimalno število stimulacijskih kanalov. Na Inštitutu za rehabilitacijo v Ljubljani so v sodelovanju s Fakulteto za elektrotehniko razvili uspešen program rehabilitacije. Kralj in Bajd [2] sta pokazala, da štirikanalna električna stimulacija zagotavlja minimalno število potrebnih stimulacijskih kanalov za hojo paraplegičnih oseb. Po dva kanala sta uporabljena za ekstenzijo kolenskega sklepa in dva za peronealno stimulacijo, ki izzove fleksijski refleks [44]. Večkanalno zaprtozančno stimulacijo je že pred desetletji uspešno pokazal Petrofsky [45]. Z razvojem regulacijske tehnike in obdelave signalov so izpopolnjeni regulatorji lahko preklapljali med hojo po ravnem terenu in po stopnicah [46]. Andrews [47] je predstavil hibridni sistem, ki je temeljil na večkanalni FES s pasivno gleženjsko opornico, regulatorjem z odločitvenimi pravili in povratno zanko. Stein [48] je nakazal uporabo električnih sistemov za izboljšanje hoje po poškodbi hrbtenjače.

1.3.2 Obnovitev vzorca hoje pri paretičnih osebah

Pri osebah, ki imajo po poškodbi hrbtenjače ohranjen del gibalnih in senzoričnih zmožnosti, je obnovitev hoje preprostejša. Poškodba hrbtenjače je prizadela živčne poti nekaterih mišičnih skupin, ki so posledično ohromljene, zato pa je hoja ovirana. Bolj poglobljena študija osrednjega živčnega sistema nam razkrije možnosti reorganizacije tega sistema, ki se je sposoben naučiti novih vzorcev hoje, pri katerih zdrave mišice, nad katerimi je nadzor ohranjen, prevzamejo nekaj nalog ohromljenih [49]. To s pridom izkoriščajo rehabilitacijski centri pri rehabilitaciji s tekočim trakom za hojo. Hoja po traku se sicer razlikuje od normalne hoje po ravnem terenu, saj se v spodnji okončini, ki zaostaja na traku (ekstenzija v kolenu), generira refleksni odziv, ki samodejno sili okončino v zamah. Da pa oseba izvede zamah, je potrebna pomoč fizioterapevta. Zaradi napornega sodelovanja fizioterapevta in težav pacienta z ravnotežjem pride do izraza sistem za razbremenitev teže. Stopnja razbremenitve je nastavljena pred terapijo glede na potrebe in poškodbo osebe. Raziskave so pokazale, da tovrstna terapija veliko doprinese k izboljšanju učinkovitosti hoje. Fizioterapevtovo prizadevanje in pomoč so nadomestili z uporabo funkcionalne električne stimulacije, kar je razbremenilo fizioterapevte in jim omogočilo lažje spremljanje pacientove hoje [50, 51, 52, 53, 54]. Tekoči trak je navadno opremljen z motorjem, v primeru da služi za analizo hoje, pa tudi s pritiskovnimi ploščami [55].

Alternativa hoji po traku in funkcionalni električni stimulaciji so naprave za hojo in robotski mehanizmi, ki se v zadnjem času zelo uveljavljajo [56]. Njihova prednost se kaže predvsem v ponovljivosti gibov, saj so ekstremitete vpete v napravo za hojo ali pa v robotski mehanizem, ki pomaga pri gibanju. V začetni fazi razvoja teh naprav so se pojavile številne pomanjkljivosti, predvsem ta, da pacient ni aktivno sodeloval v rehabilitacijskem procesu. Posledično se je izkazalo, da avtomatsko razgibavanje ni bistveno prispevalo k reorganizaciji osrednjega živčnega sistema. Hesse [57, 58] je razvil napravo, ki omogoča trening hoji podobnih gibov. Sestavljena je iz kompleksnega kolesnega mehanizma, ki omogoča nastavitev razmerja med trajanjem faze opore in faze zamaha. Vzorec hoje gibalno zdrave osebe je v njegovi napravi podoben hoji po tekočem traku. Novejša različica omogoča osebam, ki so priklenjene na invalidski voziček, vajo hoji podobnih gibov brez obremenjevanja fizioterapevta. Naprava simulira faze hoje in vodi težišče telesa v vertikalni in horizontalni smeri. V razvoju je tudi nova različica naprave s takoimenovanim haptičnim vmesnikom, ki bo omogočal sodelovanje pacienta na podlagi merjenja sil in momentov v podstavkih, na katerih pacient stoji [59]. Podoben pristop imajo na ETH v Zurichu, kjer so hojo na traku podprli z robotskim pomagalom. Colombo in ostali [60, 61] razvijajo aktivni zunanji skelet, ki omogoča premikanje spodnjih okončin pri hoji po tekočem traku. To omogočajo motorji v kolčnem in kolenskem sklepu. Med hojo pacienta robot premika ekstremitete po prednastavljenem vzorcu hoje. Novejša različica ima podajne sklepe in aktivno vodenje z merjenjem sil in momentov, kar omogoča pacientu, da do določene mere sodeluje pri rehabilitaciji. Slaba stran naprav in robotskih pomagal za hojo so težave pri zagotavljanju predpisane varnosti in predvsem zelo visoka cena.

1.4 Strategija zastavljene raziskave

Tako raziskava kakor disertacija sta zastavljeni z namenom analizirati hojo zdravih oseb in oseb z nepopolno poškodbo hrbtenjače. S tem pridobimo podatke o hoji, dobimo modele, s katerimi dopolnimo manjkajoče znanje ali načrtamo strategijo vodenja. Kinematična analiza nam omogoča izbiro alternativnih cenenih senzorjev, ki bi eventualno zadostili našim potrebam po informacijah. Za izbrane senzorje smo izdelali algoritme, ki so omogočali opis želene informacije, podajanje le-te in morebitno vodenje načrtanega sistema.

Drugi del disertacije zadeva izdelavo senzorne napravice z več senzorji, s katero smo zajemali in ocenjevali parametre hoje. Raziskati je bilo potrebno tudi možnosti povratne informacije, ki smo jo posredovali osebi med hojo. Tukaj moramo upoštevati dejstvo, da oseba med hojo ni sposobna razločiti prevelikih količin kompleksnih informacij [62], zato morajo biti posredovane informacije kar se da preproste.

Osebe, ki so utrpele poškodbo hrbtenjače in je njihovo gibanje moteno, potrebujejo pomoč pri hoji. Ohromele mišične skupine, ki jih je poškodba hrbtenjače prizadela, lahko ponovno oživimo s pomočjo FESa. Preskusili smo več načinov krmiljenja in proženja FES. Predvidena je bila uporaba enokanalne površinske stimulacije peronealnega živca, kljub temu pa predpostavimo možnost uporabe še drugega pomožnega kanala.

Posamezne segmente raziskave združimo v sistem za ponovno učenje hoje. Hojo v grobem razdelimo v dve fazi, fazo zamaha in fazo opore, zato tudi sistema načrtujemo ločeno. V disertaciji smo se posvetili fazi zamaha, medtem ko sistem za fazo opore nastaja v sodelovanju s sodelavci iz drugih držav [63].

1.5 Metodologija

Raziskavo smo začeli s kinematično analizo hoje. Hojo po ravnem terenu smo izmerili z optičnim merilnim sistemom Optotrak®, kasneje pa tudi z večsenzorno merilno napravo. Sestavljajo jo pospeškometri, žiroskop in goniometri. V disertaciji smo ugotovili, kateri senzorji nudijo zadostno informacijo za razvoj algoritmov. Ker smo želeli primerjati vrednosti izmerjene z optičnim merilnim sistemom z meritvami večsenzorne naprave, smo morali predhodno določiti še kinematični model spodnje ekstremitete. Pri tem sta bila v veliko pomoč programska paketa Matematica® in Matlab®. Po analizi hoje in delitvi v posamezne faze smo skrbno izbrali senzorje, s katerimi smo ocenili parametre hoje. Razviti algoritem [64] za ocenjevanje kvalitete zamaha je dobro merilo za vrednotenje faze zamaha.

1.5.1 Kvantitativno vrednotenje faze zamaha

Algoritem za vrednotenje kvalitete zamaha je opisan v poglavju 2.4 in predstavlja pomembni člen sistema za ponovno učenje hoje. Z njim namreč ovrednotimo zamah, s tem pa je povezano tudi učenje hoje in vodenje FESa. Algoritem potrebuje podatke iz obeh paroma pravokotnih pospeškometrov, ki merita tangencialni in radialni pospešek, in iz žiroskopa, ki meri kotno hitrost golena in goniometra, ki meri kot v kolenskem sklepu.

1.5.2 Načrtovanje vodenja motoričnega sistema in povratne informacije

Na podlagi kvalitete faze zamaha se opredelimo za najenostavnejši možni signal, ki ga lahko posredujemo osebi med hojo. V zasnovi je bil mišljen vibrotaktilni signal ali informacija v obliki električne stimulacije različnih frekvenc, vendar se je zdel pristop prekompleksen, saj bi strojna oprema ovirala osebo pri hoji po traku. Zato smo se odločili za zvočno kognitivno povratno zvezo, ki s tremi različnimi zvoki podaja kvaliteto zamaha (dober, zadovoljiv ali slab zamah).

Študije so pokazale, da lahko hoja po tekočem traku izboljša kvaliteto hoje oseb z nepopolno poškodbo hrbtenjače, vendar zahteva nenehno pomoč s strani fizioterapevtov. FES terapija s posredovanjem fizioterapevta traja kratek čas. Pri tem fizioterapevt usmerja pacienta, kdaj mora prožiti FES ali jo proži celo sam fizioterapevt, kakšno držo naj zavzame, kar se odraža v neponovljivosti in je odvisno od zavzetosti fizioterapevta. Zato bo predlagano avtomatsko proženje FESa. V ta namen je bilo potrebno izračunati še dodatno kinematično veličino, ki ne bo odvisna od precizne namestitve senzorjev in je dovolj ponovljiva. Taka veličina je kot golena, ki je bil določen z integracijo dveh senzornih skupin s Kalmanovim filtrom in je služil za proženje FESa. Na podlagi ocenjene kvalitete hoje po tekočem traku je bila osnovana regulacija jakosti stimulacije. Kolikor je oseba dosegala zahtevano kvaliteto zamahov, smo jakost stimulacije znižali, v nasprotnem primeru pa zvišali.

1.5.3 Verifikacija sistema za ponovno učenje hoje

Celoten sistem za ponovno učenje hoje je bil verificiran po korakih. S sprotnimi meritvami oseb pri hoji po traku smo lahko ugotovili pomanjkljivosti ročnega ali avtomatskega pristopa vodenja FESa. Pri tem smo želili tudi ugotoviti delovanje posameznih delov sistema, delovanje algoritma za ocenjevanje kvalitete faze zamaha, določanje kota golena, ki smo ga preverili s pomočjo optičnega merilnega sistema. Oceno kvalitete zamaha smo ocenili tudi pri meritvah brez motorične podpore, če je to dopuščala poškodba pacienta. Posameznega pacienta smo poskušali naučiti uporabe sistema za ponovno učenje hoje. Rezultati so pokazali, da bi bila izgradnja prenosnega sistema za potrebe rehabilitacijskega centra upravičena. Vse meritve in testiranja so potekala v sodelovanju in v prostorih Inštituta za rehabilitacijo Republike Slovenije.

2

Kvantitativno vrednotenje bipedalne hoje

Hoja je najobičajneši način gibanja pri človeku. Velikrat želimo tako gibanje tudi ovrednotiti, opisati. Pri tem je dobro znano dejstvo, da je hojo veliko lažje opazovati, kakor meriti. V ta namen 'opazujemo' hojo s celotno paleto video in optičnih merilnih sistemov, ki z algoritmi za obdelavo slik določajo trajektorije posameznih točk na ekstremitetah. Popolnoma drugačen pristop zahtevajo ceneni senzorji, ki merijo posamezne fizikalne komponente gibanja kot so hitrost, pospešek, kot. Razvoj in uporaba kinematike pripelje do algoritmov, s katerimi je moč dokaj natančno in enostavno opredeliti in ovrednotiti hojo. V poglavjih o kinematiki so predstavljeni osnovni pristopi, ki omogočajo izračun veličin, potrebnih za delitev hoje, v fazo zamaha in fazo opore. V nadaljevanju so opisani originalni algoritmi za vrednotenje faze zamaha, ki so lahko zaradi svoje preprostosti in uporabe cenenih senzorjev tudi osnova za aplikativne pripomočke v kineziologiji in rehabilitaciji.

2.1 Kinematika pri hoji

Analiza hoje nam daje rezultate, na podlagi katerih lahko razdelimo hojo v posamezne faze, ugotavljamo pomanjkljivosti hoje pri pacientih s poškodbo hrbtenjače in hkrati primerjamo izmerjene vrednosti z rezultati meritev drugih manj zanesljivih merilnih naprav. Namen raziskovalnega dela je bil tudi razviti večsenzorni sistem, ki nudi zadovoljivo natančnost izmerjenih podatkov. Če smo želeli izmerjene vrednosti obeh sistemov primerjati in ovrednotiti, je bilo potrebno izračunati skupne kinematične veličine. Kot referenčni merilni sistem je bil uporabljen optični merilni sistem Optotrak®. Delovanje optičnega merilnega sistema temelji na uporabi markerjev, infrardečih LED diod, ki jih pritrdimo na gibajoče se telo, položaj markerjev v prostoru pa zaznavamo z infrardečimi kamerami. Markerje smo namestili v metatarzalnem sklepu, na peti in v sklepih v gleženju, kolenu in kolku (T_1, T_2, T_3, T_4, T_5). S pomočjo vektorske analize in transformacij koordinatnih sistemov lahko iz položaja markerjev izračunamo vse ostale veličine, ki jih potrebujemo za analizo hoje in vrednotenje prenosnega večsenzornega sistema.

Vsakemu markerju je prirejen krajevni vektor iz baznega koordinatnega sistema ($\vec{T_1} = [T_{1x}, T_{1y}, T_{1z}]$...). Iz krajevnih vektorjev določimo vektorje, ki so vzporedni s segmenti spodnje ekstremitete, v stopalu, golenu in stegnu:

$$\begin{aligned} \vec{T}_{21} &= \vec{T}_2 - \vec{T}_1 \\ \vec{T}_{43} &= \vec{T}_4 - \vec{T}_3 \\ \vec{T}_{54} &= \vec{T}_5 - \vec{T}_4 \end{aligned} \tag{2.1}$$

S skalarnim produktom in kosinusnim izrekom izračunamo kota v gleženjskem in kolenskem sklepu:

$$\varphi_0 = \arccos \frac{\vec{T}_{43} \cdot \vec{T}_{21}}{|\vec{T}_{43}||\vec{T}_{21}|} \tag{2.2}$$

$$\varphi_1 = \arccos \frac{\vec{T}_{54} \cdot \vec{T}_{43}}{|\vec{T}_{54}||\vec{T}_{43}|} \tag{2.3}$$

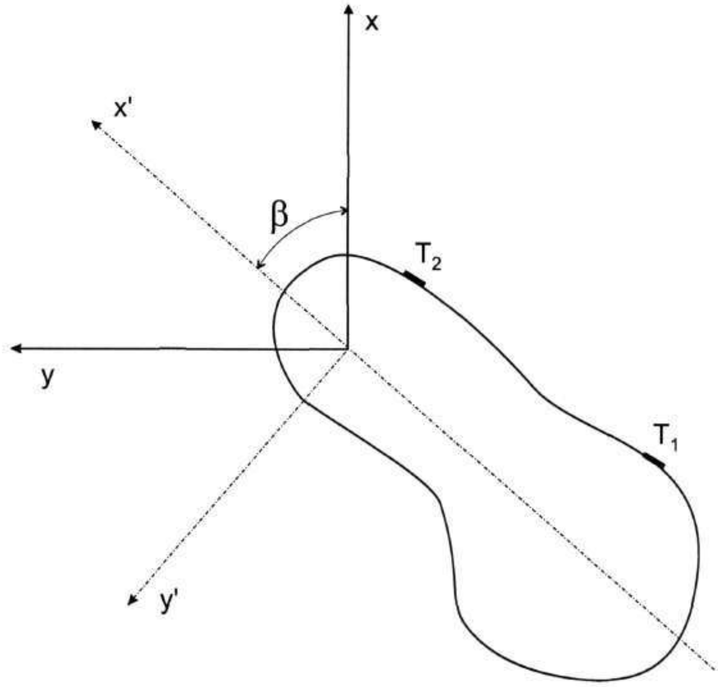
Izračunani kinematični veličini lahko uporabimo za vrednotenje hoje [44], krmiljenje električne stimulacije ali primerjamo z izmerjenimi vrednostmi prenosne večsenzorne naprave (slika 2.3), s katero smo želeli nadomestiti optični merilni sistem, katerega uporaba je zaradi visoke cene in prostorskih zahtev omejena na laboratorij.

Bolj zahtevno je vrednotenje izmerjenega pospeška. Če želimo primerjati izmerjene vrednosti, jih je potrebno preračunati na skupno veličino. Izbrana skupna veličina je bila pospešek v gleženjskem sklepu. Po enačbi 2.20 izračunamo pospešek gleženskega sklepa iz merjenih vrednosti večsenzorne naprave, medtem ko moramo pri izračunu iz krajevnih vektorjev markerjev upoštevati tudi eksterno kolensko rotacijo in posledično stopala β (slika 2.1):

$$\begin{bmatrix} x' \\ y' \\ z' \end{bmatrix} = \begin{bmatrix} \cos(\beta) & \sin(\beta) & 0 \\ -\sin(\beta) & \cos(\beta) & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} x \\ y \\ z \end{bmatrix} \tag{2.4}$$

$$\beta = \arccos \frac{\vec{T}_{210} \cdot \vec{T}_{21r}}{|\vec{T}_{210}||\vec{T}_{21r}|} \tag{2.5}$$

kjer sta $\vec{T}_{210} = [T_{2x} - T_{1x}, 0, 0]$ in $\vec{T}_{21r} = [T_{2x} - T_{1x}, T_{2y} - T_{1y}, 0]$ vektorja.



Slika 2.1: Slika prikazuje zasuk koordinatnega sistema stopala za kot β proti koordinatnemu sistemu v sklepu gležnja. To je posledica eksterne rotacije [44] pri zamahu.

Kot med podlago in stopalom α izračunamo iz markerjev 1 in 2:

$$\alpha = \arctan \frac{T_{2z} - T_{1z}}{|\vec{T}_{21}|} \tag{2.6}$$

Večsenzorni sistem je pritrjen na goleno in se giblje v skladu s položajem tibie. Ker pa so rezultati iz optičnega merilnega sistema za gibanje v sagitalni ravnini, pride pri eksterni rotaciji pri zamahu do razlik. Zato uporabimo kot β za transformacijo gibanja iz sagitalne ravnine (XY) v ravnino večsenzorne naprave (X'Y')(slika 2.1):

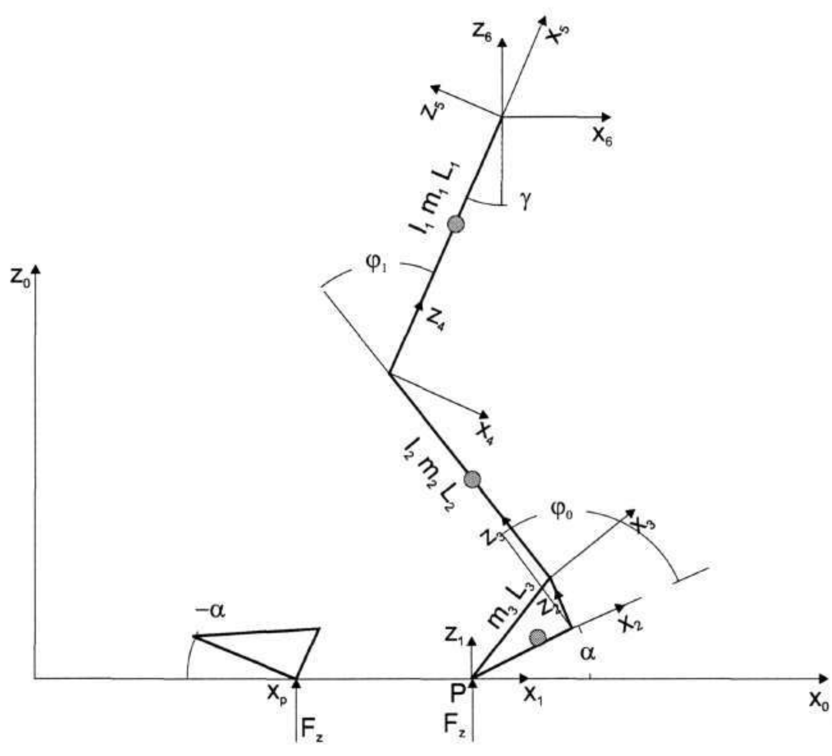
$$\begin{bmatrix} x_g \\ y_g \\ z_g \end{bmatrix} = \begin{bmatrix} \cos(\beta) & \sin(\beta) & 0 \\ -\sin(\beta) & \cos(\beta) & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} T_{3x} \\ T_{3y} \\ T_{3z} \end{bmatrix} \tag{2.7}$$

Pozicijo gleženjskega sklepa, transformirano v ravnino X'Y' (x_g, z_g) numerično odvajamo s formulo za dvakratno odvajanje z enačbo 2.8, kjer je T_s vzorčni čas:

$$\begin{aligned} \ddot{x}_g(i) &= \frac{-x_g(i+3) + 4x_g(i+2) - 5x_g(i+1) + 2x_g(i)}{T_s^2} \\ \ddot{z}_g(i) &= \frac{-z_g(i+3) + 4z_g(i+2) - 5z_g(i+1) + 2z_g(i)}{T_s^2} \end{aligned} \tag{2.8}$$

Dvakratni odvod vnaša v sistem precej šuma, zato signal filtriramo z nizkoprepustnim filtrom. Nato dobljeni pospešek (x_g, z_g) ob upoštevanju gravitacije g transformiramo iz ravnine večsenzorne naprave v gibanje v tangencialno - radialni smeri glede na goleno (enačba 2.9):

$$\begin{aligned} a_{t0opt} &= -\ddot{x}_g \cdot \sin(\varphi_0 + \alpha) + (g + \ddot{z}_g) \cdot \cos(\varphi_0 + \alpha) \\ a_{r0opt} &= -\ddot{x}_g \cdot \cos(\varphi_0 + \alpha) - (g + \ddot{z}_g) \cdot \sin(\varphi_0 + \alpha) \end{aligned} \tag{2.9}$$

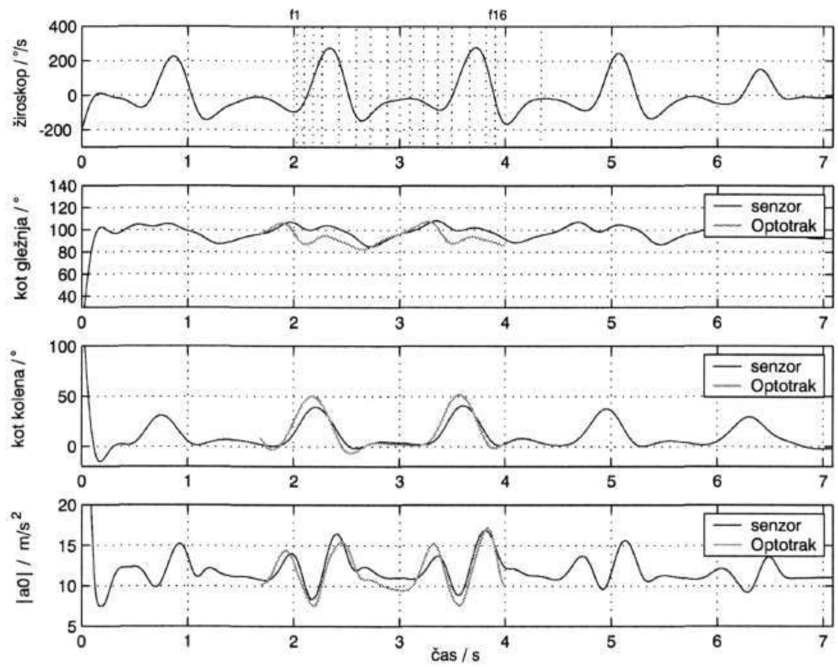


Slika 2.2: Postavitev koordinatnih sistemov v modelu spodnje okončine. V koordinatnih sistemih v kolenu, gležnju, kolku, na peti in v metatarzalnem sklepu so pritrjeni markerji optičnega merilnega sistema. Izračunamo kinematične veličine φ_0, φ_1, α in položaje težišč, ki jih potrebujemo za analizo hoje. Prijemališče x_p zunanje sile F služi za analizo faze opore.

Glede na sliko 2.2 in koordinatni sistem 3 lahko zapišemo enačbo 2.10:

$$\begin{aligned} a_{t0opt} &= -\ddot{x}_3 \\ a_{r0opt} &= -\ddot{z}_3 \\ |a_{0opt}| &= \sqrt{a_{t0opt}^2 + a_{r0opt}^2} \end{aligned} \tag{2.10}$$

Izračun potrebnih kinematičnih veličin z optičnim merilnim sistemom nam je omogočil vrednotenje izmerjenih vrednosti z večsenzorno napravo (slika 2.3). V kolenskem in gleženjskem sklepu sta bila nameščena goniometra za merjenje kotov φ_0 in φ_1. Iz merjenih vrednosti tangencialnih in radialnih pospeškov s pospeškometri je bil določen absolutni pospešek gleženjskega sklepa $|a_0|$ (enačba 2.20).



Slika 2.3: Slika prikazuje primerjavo veličin, izračunanih iz izmerjenih vrednosti Optotraka® in večsenzorne naprave. Iz zajetih signalov lahko razberemo značilke v posameznih signalih, ki določajo faze zamaha ali opore. Razvita programska oprema v okolju Matlab nam omogoča, da izbrano točko na grafu (razvidno v sledi žiroskopa od f1 do f16) prikažemo na modelu noge (slika 2.4).

S kinematično analizo z optičnim merilnim sistemom Optotrak® smo pokazali sprejemljivo natančnost večsenzornega sistema in hkrati potrebne kinematične veličine za opis gibanja spodnje ekstremitete. Tukaj imamo v mislih tangencialni in radialni pospešek gleženjskega sklepa, ki omogoča kvantitativno vrednotenje hoje [64], določanje faze zamaha in opore, merjeno kotno hitrost golena in kote v kolenskem in gleženjskem sklepu.

Merili smo tudi sile in momente v fazi opore, ko je oseba stopila na pritiskovno ploščo in izračunali prijemališče dotične sile. V nadaljevanju disertacije smo se osredotočili na fazo zamaha, medtem ko analiza faze opore in razvoj pripadajočega sistema ostaja predmet skupnih raziskav s sodelavci v tujini [63].

2.2 Delitev hoje v posamezne faze

V grobem razdelimo hojo v dve fazi, fazo zamaha in fazo opore. Faza zamaha se nanaša na stanje, ko je ena noga dvignjena, medtem ko je faza opore določena z dotikom noge s tlemi. Za začetek faze zamaha štejemo dvig prstov s tal, za konec in hkrati začetek faze opore pa dotik s peto [44]. Časovno razmerje med fazo zamaha in fazo opore pri normalni hoji je 40:60. Hitrejša hoja proporcionalno podaljša enojno fazo opore (stik s tlemi z eno nogo) in skrajša interval dvojne opore (stik s tlemi z obema nogama). Vsako fazo, torej fazo zamaha in opore, lahko razdelimo še v osem funkcinalnih delov, podfaz ali funkcionalnih intervalov.

Fazo opore v:

- začetni dotik,
- obremenitev,
- osrednjo fazo opore,
- končno fazo opore.

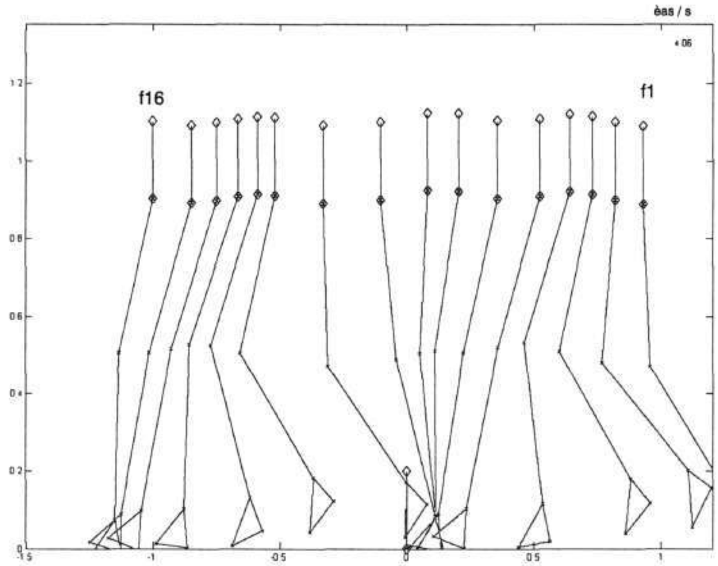
Fazo zamaha pa v:

- predzamah,
- začetno fazo zamaha,
- osrednjo fazo zamaha,
- končno fazo zamaha.

Delitev velja tako za normalno, kakor patološko hojo. Pri slednji so sicer posamezne faze neizrazite, ker gre za motnje v gibanju, ki so posledica možganske kapi, poškodbe hrbtenjače, multiple skleroze ali artritisa. Na podlagi meritev in delitve v faze lahko natančneje določimo tip patološke hoje in ustrezno terapijo [44].

2.2.1 Prikaz posameznih faz preko meritev

Kinematika spodnje ekstremitete je bila določena v prejšnjem poglavju z optičnim merilnim sistemom Optotrak®. Da bi lahko prikazal in analiziral posamezne podfaze pri hoji, sem uporabil kinematične podatke, ki so bili na voljo (slika 2.3). Izdelana programska oprema je omogočala sledenje izmerjenim kinematičnim podatkom na sliki (slika 2.4), kar je bilo v izredno pomoč pri določanju značilk za samodejno zaznavanje faze zamaha, opore in analizo hoje.



Slika 2.4: Prikaz izmerjenih kinematičnih vrednosti v modelu.

2.3 Določanje faz hoje z umetnimi senzorji

Potem, ko je hoja razdeljena v posamezne faze, je smiselno razmisliti o načinu določanja oz merjenja posameznih faz. Vsaka faza hoje ima svoje značilke v merjenih signalih, ki jih zaznamo s specifičnimi merilnimi pristopi. Zaradi raznolikosti značilk pride do izraza izbira merilnega sredstva, senzorja. Pri uporabi optičnega merilnega sistema dobimo položaj ekstremitet, s pomočjo izračunov pa še ostale veličine, kot so translacijske in kotne hitrosti ter pospeški. Velikokrat pa optični merilni sistem ni na voljo ali njegova uporaba ni smiselna, zato se poslužujemo cenejših sistemov, osnovanih na cenenih senzorjih, pospeškometrih, žiroskopih, merilnikih tresljajev, goniometrih, inklinometrih. Pri tem si brez uporabe dobrih algoritmov ne moremo obetati uspešnega razpoznavanja značilk. V disertaciji bosta predstavljena dva načina razpoznavanja faze zamaha z dvema skupinama senzorjev, pospeškometri ali žiroskopom. Eden izmed načinov je z uporabo nevronskih mrež, drug precej preprostejši uporablja izpopolnjeno sekantno metodo.

2.3.1 Uporaba nevronskih mrež

Fizikalni model lahko predstavimo z matematičnimi enačbami ali tabelami, katerih parametre določimo z meritvijo ali pa so že predhodno določeni kot zahteve procesa. Posamezne parametre, ki jih ni moč določiti z meritvijo, določimo naknadno z meritvijo vhodno-izhodne karakteristike, ki je osnova za identifikacijo želenih parametrov. Tretji primer pa nastopi, ko pravzaprav nimamo nobenih informacij o fizikalnem ozadju procesa, nastopi pa potreba po modeliranju. Takrat se poslužujemo identifikacijskih metod, ki temeljijo, na primer, na učenju umetnih nevronskih mrež. Nevronske mreže sem uporabil za določanje faze zamaha v realnem času iz izmerjenih signalov pospeškometrov ali žiroskopa (poglavje 2.3.2).

Nevronske mreže sestavljajo enote, ki imajo vhod, ki je utežen z utežjo, prednastavitev (bias) in izhod. Slednji je definiran z aktivacijsko funkcijo [65]. Enote posameznega nivoja med sabo niso povezane, povezave so možne le med nivoji. Tako ločimo dvonivojske mreže, ki imajo en skriti nivo in večnivojske mreže, ki jih imajo več. Glede na tip povezav pa razlikujemo 'feedforward' mreže, ki nimajo povratnih vezav, torej izhod ni odvisen od prejšnjih vrednosti izhodov, in 'recurrent' mreže, ki vsebujejo povratne povezave.

Identifikacija dinamičnega modela temelji na vhodno-izhodnem opisu sistema. Stanja sistema vzbudimo z vhodnimi signali u(t) in opazujemo njegove odzive y(t), ki jih uporabimo za učenje nevronske mreže:

$$\begin{aligned} u(t) &= [u_1, u_2 ... u_k(t)] \\ y(t) &= [y_1, y_2 ... y_k(t)] \end{aligned} \tag{2.11}$$

V enačbi 2.11 obstajajo povezave med preteklimi vrednostmi vhoda in izhoda in prihodnjimi izhodi. To izrazimo s funkcijo g, ki jo poiščemo v družini funkcij. Če družino funkcij parametriziramo v vektorski obliki, lahko zapišemo:

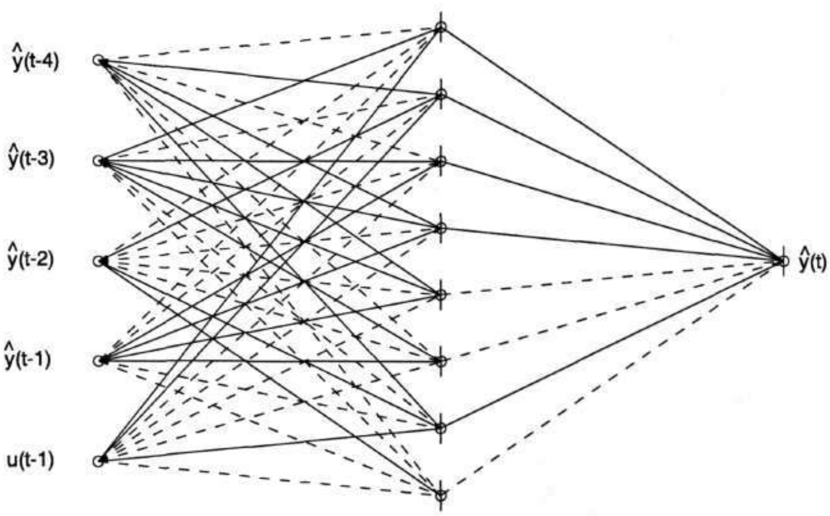
$$g(u^{t-1}, y^{t-1}, \Phi) = g(\varphi(t), \Phi), \tag{2.12}$$

kjer je $\varphi(t) = \varphi(u^{t-1}, y^{t-1})$ regresijski vektor. Da zadostimo enačbi 2.12, moramo rešiti dva problema, določiti regresijski vektor $\varphi(t)$ iz preteklih vhodov in izhodov ter izbrati nelinearno preslikavo $g(\varphi)$ iz regresijskega prostora v prostor izhodov. Z eksperimentom določimo vhodno-izhodne lastnosti dinamičnega sistema. Pri tem uporabimo ustrezni vhodni signal, katerega frekvenčni spekter vzbudi vsa stanja sistema. V naslednjem koraku izberemo strukturo modela, torej model nevronske mreže, število osnovnih enot, ki sestavljajo mrežo ter določimo regresorje. Iz niza podatkov 2.13 moramo dobiti možne uteži nevronske mreže:

$$\begin{gathered} Z^N = \{[u(t), \varphi(t)], t = 1..N\} \\ Z^N \to \Phi \end{gathered} \tag{2.13}$$

Uteži Φ so izbrane tako, da je izhod nevronske mreže $\hat{y}(t)$ čim bolj enak pravemu izhodu $y(t)$. Za oceno možnih uteži minimiziramo kriterijsko funkcijo:

$$min_\Phi V_N(\Phi, Z^N) = \frac{1}{N}\sum_{t=1}^{N} \|y(t) - g(\varphi(t), \Phi)\|^2 \tag{2.14}$$



Slika 2.5: Primer dvonivojske nevronske mreže četrtega reda z osmimi skritimi enotami in enim izhodom.

Uteži so bile optimizirane z Levenberg-Marquardtovo metodo [66]. Pri izbiri strukture modela (slika 2.5) je bila v pomoč simulacija dvonivojske nevronske mreže z n enotami. Izmed vseh je bila izbrana najenostavnejša, ki je še zadovoljivo opisovala obnašanje sistema in uporablja pretekle vhode $u(t-k)$ in pretekle izhode $y(t-k)$ kot regresorje:

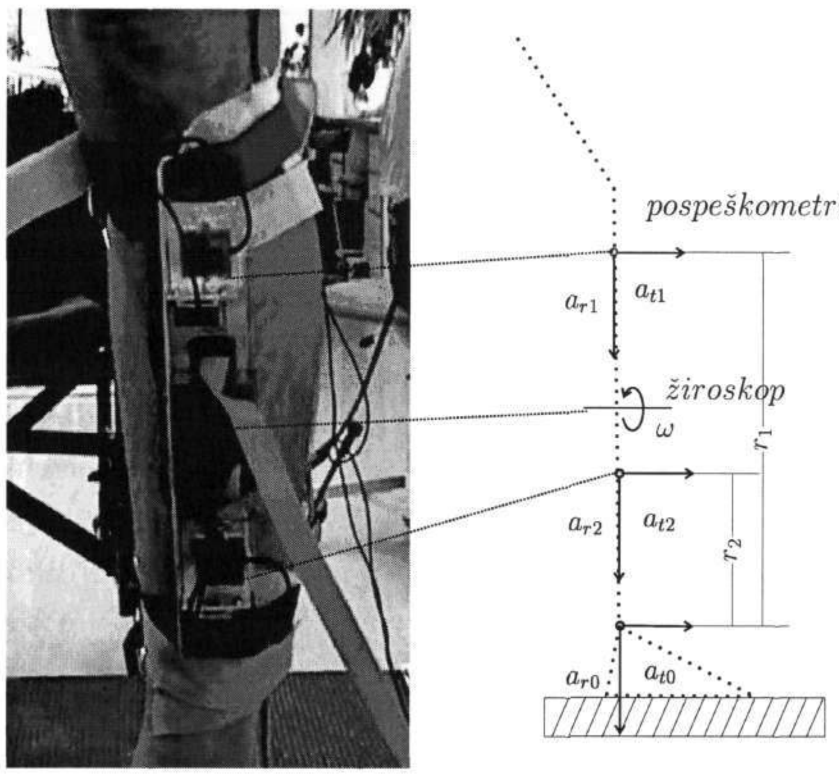
$$\varphi(t) = [y(t-1)...y(t-n_a), u(t-n_k)...u(t-n_b-n_k+1)], \tag{2.15}$$

kjer je n_a število preteklih izhodov, n_b število preteklih vhodov, ki vplivajo na trenutni izhod, in n_k zakasnitev sistema. In kot prediktor uporabimo:

$$\hat{y}(t|\Phi) = \hat{y}(t|t-1, \Phi) = g(\varphi(t), \Phi) \tag{2.16}$$

2.3.2 Uporaba pospeškometrov in žiroskopa

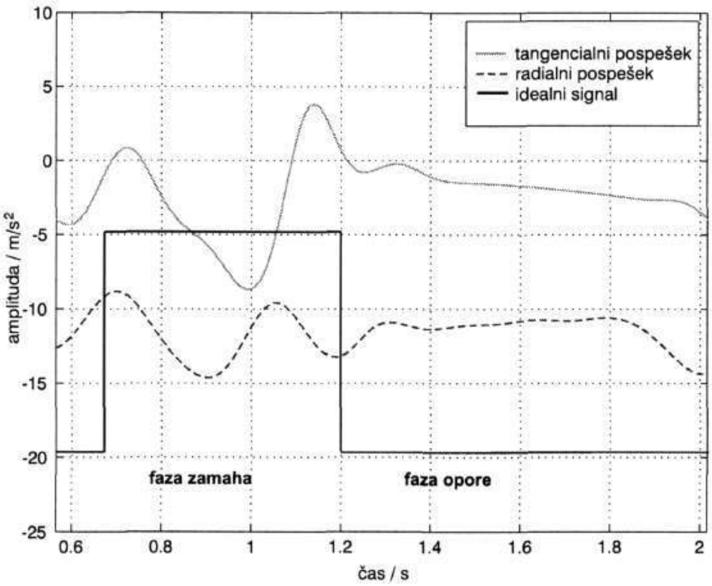
Z uporabo opisanega algoritma lahko naučimo nevronsko mrežo oziroma določimo takšne uteži, ki bodo vhodni signal preoblikovale v želen izhodni signal. Kot vhodni signal lahko uporabimo zaznane veličine, kot v kolenu, gležnju, pospeške v gleženjskem sklepu ali kotno hitrost golena, morda pa kombinacijo posameznih ali vseh signalov. Kot vhodni signal za učenje nevronske mreže sem izbral kombinacijo obeh pospeškov v gleženjskem sklepu.



Slika 2.6: Prva različica večsenzorne naprave, opremljene z dvema paroma pospeskometrov in žiroskopom, je nameščena na golenu osebe. V kolenskem in gleženjskem sklepu pa sta nameščena goniometra za merjenje kota vsakega sklepa. Na desni je prikazan diagram pospeškov, kotne hitrosti in shematska razlaga koordinatnih sistemov v sagitalni ravnini.

Za potrebe razpoznavanja faze zamaha pri hoji je bilo potrebno določiti pospeška v gleženjskem sklepu. V tem sklepu dobimo namreč najbolj razgibane trajektorije, ki jih lahko primerjamo s kinematičnimi meritvami z optičnim merilnim sistemom. Pospešek v gleženjskem sklepu določimo z dvema paroma pospeškometrov, nameščenih na golenu osebe tako, da paroma merijo tangencialno in radialno komponento pospeška (slika 2.6).

Za vsako izhodišče obeh koordinatnih sistemov pospeškometrov lahko zapišemo



Slika 2.7: Pospeška v gležnju kot vhodni signal za učenje nevronske mreže in idealni izhodni signal, ki določa fazo zamaha. Idealni signal je določen na podlagi meritev iz poglavja 2.2.1 in literature [44].

enačbo:

$$\begin{bmatrix} x \\ z \end{bmatrix} = \begin{bmatrix} x_0 \\ z_0 \end{bmatrix} + r \cdot \begin{bmatrix} cos\vartheta \\ sin\vartheta \end{bmatrix}, \tag{2.17}$$

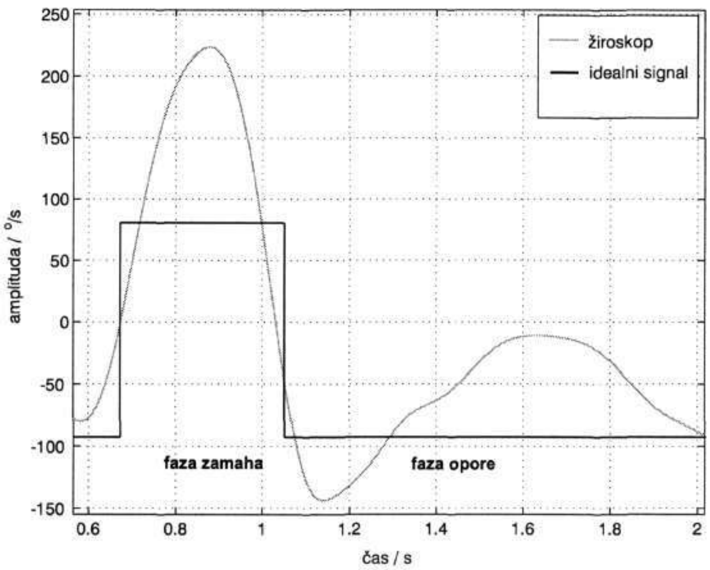
kjer sta x_0 in z_0 koordinati gleženjskega sklepa ter ϑ kot med golenom in horizontalno ravnino. Če enačbo 2.17 dvakrat odvajamo in preoblikujemo, dobimo izražen pospešek koordinatnega sistema prvega para pospeškometrov $\mathbf{a}_1$ v enačbi pa nastopa $\mathbf{a}_0$ kot pospešek koordinatnega sistema gleženjskega sklepa:

$$\mathbf{a}_1 = \mathbf{a}_0 + r_1\ddot{\vartheta} \cdot \begin{bmatrix} -sin\vartheta \\ cos\vartheta \end{bmatrix} - r_1\dot{\vartheta}^2 \cdot \begin{bmatrix} cos\vartheta \\ sin\vartheta \end{bmatrix} \tag{2.18}$$

Pospeške posameznih koordinatnih sistemov zapišemo v vektorski obliki:

$$\mathbf{a}_1 = \begin{bmatrix} a_{t1} \\ a_{r1} \end{bmatrix} \quad \mathbf{a}_2 = \begin{bmatrix} a_{t2} \\ a_{r2} \end{bmatrix} \tag{2.19}$$

in jih vstavimo v enačbo 2.18. Enak postopek naredimo za drug par pospeškometrov in dobimo še $\mathbf{a}_2$. Tako dobimo dve enačbi za oba para pospeškometrov. r_1 in r_2 sta razdalji med gleženjskim sklepom in posameznima paroma pospeškometrov. Če enačbo 2.18 pomnožimo z r_1, enačbo za drug par pa z r_2 ter ju odštejemo, lahko izrazimo pospešek gleženjskega sklepa (slika 2.6) [22]:



Slika 2.8: Kotna hitrost golena kot vhodni signal za učenje nevronske mreže in idealni izhodni signal, ki določa fazo zamaha. Idealni signal je določen na podlagi meritev iz poglavja 2.2.1 in literature [44].

$$\mathbf{a}_0 = \begin{bmatrix} a_{t0} \\ a_{r0} \end{bmatrix} = \frac{1}{r_1 - r_2}(r_1 \cdot \begin{bmatrix} a_{t2} \\ a_{r2} \end{bmatrix} - r_2 \cdot \begin{bmatrix} a_{t1} \\ a_{r1} \end{bmatrix}) \tag{2.20}$$

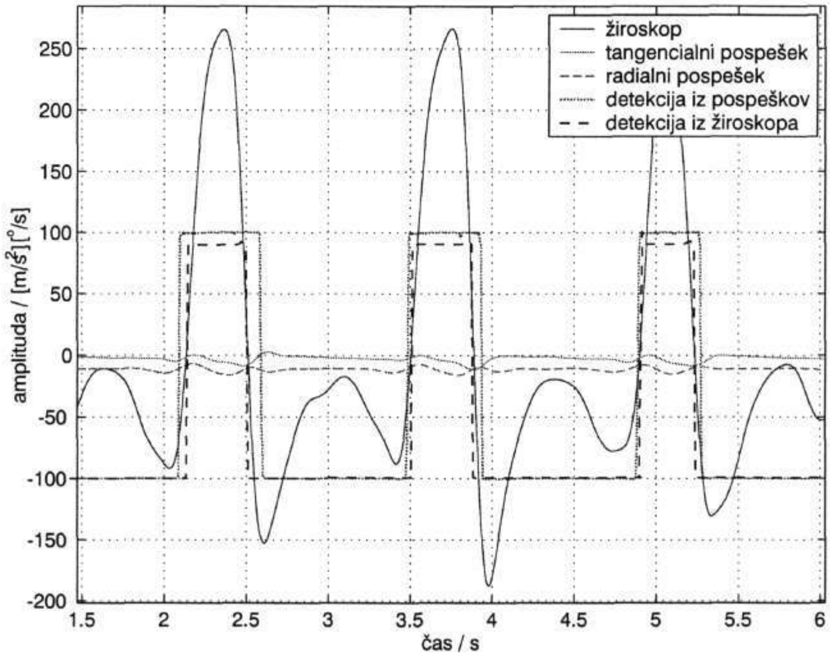
Veličine določene z enačbo 2.20 uporabimo kot vhodni signal za učenje nevronske mreže, izhodni signal pa izberemo glede na meritve v poglavju 2.2.1 in glede na poznavanje goniogramov [44] pri fazi zamaha oz opore (slika 2.7 in 2.8). Strukturo nevronske mreže določimo eksperimentalno, tako da ima najmanjše število n_a preteklih izhodov in preteklih vhodov ter skritih nevronov (tabela 2.1).

Tabela 2.1: Parametri nevronske mreže za razpoznavanje faz s pospeškometri oziroma žiroskopom.

Tip vhodnega signala	Število skritih nevronov	n_b	n_a
pospeškometer	10	[2 2]	1
žiroskop	7	[1]	1

Dvonivojska (1 skrit nivo in 1 izhodni nivo nevronov) nevronska mreža (slika 2.5) je enostaven, vendar učinkovit pristop, ki ga realiziramo v programskem paketu *Simulink* / *Matlab*® [66].

Preizkušanje zanesljivosti določanja faze zamaha je potekalo s prvo prototipno napravo (slika 2.6) na treh zdravih in eni paretični osebi pri hoji na ravnem terenu. Na



Slika 2.9: Izhodni signali naučene dvonivojske nevronske mreže (tabela 2.1). Vhodni signali (pospeškometri ali žiroskop) pri učenju nevronske mreže in izhodna signala nevronske mreže, ki določata fazo zamaha.

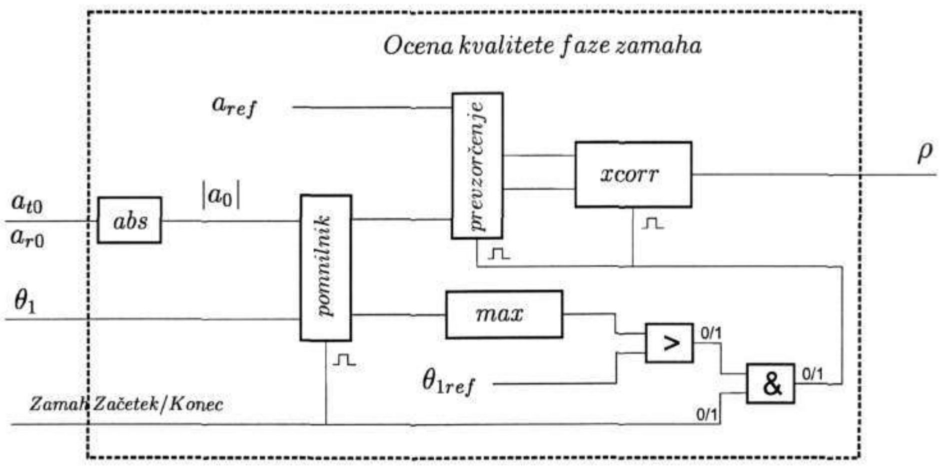
sliki 2.9 so prikazani rezultati določanja faze zamaha. Obsežne meritve so pokazale, da določanje faze zamaha s signali pospeškometrov ni zanesljivo. Razlog je bila predvsem različna obutev, kar bistveno vpliva na pospeške pri dostopu. Tako je v nadaljnjih poglavjih pri določanju faze zamaha uporabljena metoda detekcije faze zamaha z žiroskopom.

Možno je uporabiti [67, 64] tudi drugo metodo določanja faze zamaha. Izhodni signal žiroskopa je pravzaprav kotna hitrost golena. Le-ta spremeni predznak ob spremembi smeri gibanja ekstremitete. Torej je spremembo možno izkoristiti za detekcijo faze zamaha. Res je možno, da se pojavijo spremembe predznaka tudi v fazi opore [64], vendar teh ne upoštevamo pri nadaljnji uporabi, saj jih je možno izločiti na podlagi analize trajektorije kateregakoli zajetega signala v fazi zamaha.

2.4 Kvantitativna ocena faze zamaha

V prejšnjih poglavjih je predstavljena delitev hoje v posamezne faze. V tem poglavju se bom osredotočil na fazo zamaha, ki je v novem originalnem algoritmu merodajna za novonastajajoč pristop k ponovnemu učenju hoje. Algoritem temelji na shranjevanju zajetih podatkov v času faze zamaha. Izmerjene podatke pospeškometrov in kolenski goniogram shranimo v začasni pomnilnik. Glede na zahteve fizioterapevta in gibalne sposobnosti pacienta nastavimo tudi referenčne vrednosti, zahtevani kot θ_{1ref} v kolenu med zamahom in referenčno trajektorijo pospeška.

Tako referenčno trajektorijo, kakor trajektorijo pospeška med hojo tvorita oba para pospeškometrov, iz katerih določimo pospešek gleženjskega sklepa (enačbe 2.17 do 2.20). Izračunana veličina je ključnega pomena pri ocenjevanju kvalitete faze zamaha. Algoritem (slika 2.10) temelji na predhodnem zajetju trajektorije pospeška v gleženjskem sklepu, ki potem služi kot referenca za ocenjevanje. V večini primerov je pri osebah z nepopolno poškodbo hrbtenjače trajektorija pospeška manj prizadete spodnje ekstremitete. Referečno vrednost kota v kolenu θ_{1ref} nastavimo glede na gibalne sposobnosti osebe in predstavlja minimalno fleksijo v kolenu, ki jo mora oseba doseči pri zamahu, da lahko govorimo o zadovoljivem zamahu. Če pogoj ni izpolnjen, se ocena kvalitete faze zamaha zavrže, zamah pa označi z nezadostnim. V primeru izpolnjevanja pogoja kolenske fleksije se izvede algoritem za ocenjevanje kvalitete faze zamaha (slika 2.10) [64].



Slika 2.10: Shema algoritma za ocenjevanje kvalitete faze zamaha.

Zajemanje signalov se prične, ko nastopi faza zamaha. Slednjo zaznamo z algoritmom za detekcijo faze zamaha (poglavje 2.3.2). Med fazo zamaha shranjujemo podatke o absolutni vrednosti pospeška $|\mathbf{a}_0|$ in o fleksiji kolena v pomnilnik. Takoj po detekciji konca faze zamaha se ob izpolnjevanju že omenjenega pogoja fleksije kolena izvede prevzorčenje shranjene trajektorije pospeška $|\mathbf{a}_0|$. Zahteva po prevzorčenju signala nastopi zaradi različne dolžine oz hitrosti zamaha in s tem posledično števila vzorcev.

Prevzorčen signal je nato normiran in koreliran (enačba 2.21) z referenčno trajektorijo $|\mathbf{a}_{ref}|$.

$$\varphi_{0,r}(kT,\tau) = \frac{1}{n}\sum_{k=1}^{n}|a_0(kT)|\cdot|a_{ref}(kT+\tau)| = E[|a_0(kT)|\cdot|a_{ref}(kT+\tau)|] \quad (2.21)$$

$$\rho_{0,r} = \frac{E[(a_0(kT)-m_0(kT))\cdot(a_{ref}(kT+\tau)-m_{ref}(kT+\tau))]}{\sqrt{E[(a_0(kT)-m_0(kT))^2\cdot(a_{ref}(kT+\tau)-m_{ref}(kT+\tau))^2]}} \quad (2.22)$$

V zgornji enačbi je m_0 srednja vrednost merjenega signala, m_{ref} pa srednja vrednost referenčnega pospeška. Korelacijski faktor $\rho_{0,r}$ (enačba 2.22) je hkrati tudi merilo za kvaliteto faze zamaha. Če je enak 0, potem se signala ne ujemata in je zamah slab, z naraščanjem proti vrednosti 1 se izboljšuje in postane popoln.

2.5 Uporaba Kalmanovega filtra

Sisteme, v katerih uporabimo več senzorjev za zajemanje iste ali posredno povezane informacije, lahko imenujemo kompleksne sisteme. Algoritmi, ki procesirajo informacije v takih sistemih, še posebej če je tak sistem v zaprtozančnem procesu, postanejo zapleteni, predvsem pa računsko prezahtevni za izvedbo v realnem času. Zmanjšanje števila senzorjev ali poenostavitve pa včasih vodijo do zmanjšanja zanesljivosti delovanja. Zato velja kompleksne sisteme decentralizirati, razbiti na več manjših podsistemov. To potegne za sabo veliko prednosti. Razbit, decentraliziran, multisenzorni sistem ima več senzorjev, ki so lahko komplementarni, dopolnjujoči [68] ali redundantni [69], kar povečuje zanesljivost sistema in dopolnjuje manjkajoče informacije za nazornejši prikaz merjene veličine. Včasih je decentralizacija neizbežna zaradi prostorske razporeditve informacij in posledično tudi senzorjev.

Informacijo multisenzornih sistemov je potrebno združiti, da predstavlja razumljivo, zanesljivo in predstavljivo informacijo. Principi združevanja, fuzije ali integracije senzornih informacij temeljijo na teoriji Kalmanovega filtra [70].

Najpogostejšo uporabo senzorne integracije najdemo v navigaciji, letalski, vesoljski tehniki, robotiki in procesnih regulacijah. Integrirani navigacijski sistemi (INS), ki jih sestavljajo žiroskopi in pospeškometri so najlepši primer uporabe omenjene tehnike. Prav tako gre pri opremi za merjenje razdalje (distance measuring equipment, DME) za določanje veličin, ki jih ne moremo neposredno meriti.

2.5.1 Teorija Kalmanovega filtra

Leta 1960 je R.E.Kalman objavil svoj najbolj znani članek [70], ki opisuje rekurzivno reševanje diskretnega linearnega filtriranja podatkov, vendar je teorija obvisela v zraku vse do nastopa digitalnih računalnikov. Predstavlja skupek matematičnih enačb za učinkovito numerično reševanje metode najmanjših kvadratov.

Kalmanov filter je rekurzivni linearni estimator, ki sukcesivno izračuna minimalno varianco za stanja v času na podlagi periodičnega opazovanja, ki je linearno odvisno od tega stanja. Kalmanov filter minimizira srednjo kvadratično napako in je optimalen glede na razne kriterije v procesu in glede ne šum. Razvoj linearnih estimatorjev lahko razširimo tudi na problem ocenjevanja nelinernih sistemov [71].

Kalmanov filter ocenjuje stanja sistema ob upoštevanju znanih vhodov ter množice izmerjenih signalov. Sistem v diskretni obliki, ki upošteva tudi signal šuma na vhodu sistema $\mathbf{w}_k$ ter šum senzorjev $\mathbf{v}_k$, opisuje model stanj [72]

$$\begin{aligned} \mathbf{x}_{k+1} &= \phi_k \mathbf{x}_k + \mathbf{B}\mathbf{u}_k + \mathbf{w}_k \\ \mathbf{z}_k &= \mathbf{H}_k \mathbf{x}_k + \mathbf{v}_k \,, \end{aligned} \tag{2.23}$$

kjer vektor $\mathbf{z}_k$ predstavlja vektor vseh merjenih signalov, ki so na voljo za Kalmanov filter, ϕ matriko prehajanja stanj, B vhodno matriko in H izhodno matriko. Predpostavimo, da sta šum sistema ter merilni šum bela šuma s kovariančnima matrikama $\mathsf{Q}_{\mathbf{w}k}$ ter $\mathsf{R}_{\mathbf{v}k}$

$$E[\mathbf{w}_k \mathbf{w}_i^T] = \begin{cases} \mathsf{Q}_{\mathbf{w}k} & : \; i = k \\ 0 & : \; i \neq k \end{cases} \tag{2.24}$$

$$E[\mathbf{v}_k \mathbf{v}_i^T] = \begin{cases} \mathsf{R}_{\mathbf{v}k} & : \; i = k \\ 0 & : \; i \neq k \end{cases} \tag{2.25}$$

$$E[\mathbf{w}_k \mathbf{v}_i^T] = 0, \quad za\ vse \quad k \quad in \quad i \tag{2.26}$$

Predpostavljamo, da imamo ob času t_k definirano začetno oceno procesa. Ta ocena je 'a priori' ocena $\hat{\mathbf{x}}_k^-$, kjer minus označuje najboljšo oceno ob trenutku t_k. Predpostavimo tudi, da poznamo kovariančno matriko povezano z $\hat{\mathbf{x}}_k^-$. Definiramo napako ocene:

$$e_k^- = \hat{\mathbf{x}}_k - \hat{\mathbf{x}}_k^- \tag{2.27}$$

in spremljajočo kovariančno matriko napake:

$$\mathbf{P}_k^- = E[\mathbf{e}_k^- \mathbf{e}_k^{-T}] = E[(\hat{\mathbf{x}}_k - \hat{\mathbf{x}}_k^-)(\hat{\mathbf{x}}_k - \hat{\mathbf{x}}_k^-)^T] \tag{2.28}$$

Kalmanov filter generira oceno stanj sistema, ki minimizira srednjo kvadratično napako ocene (enačba 2.28). Pri izračunu enačb za Kalmanov filter ponavadi začnemo z iskanjem enačbe, ki izraža a posteriori stanje, kot linearno kombinacijo a priori stanja in utežene razlike med dejansko meritvijo $\mathbf{z}_k$ in predikcijo meritve $\mathbf{H_k} \cdot \hat{\mathbf{x}}_k^-$, kakor prikazuje enačba 2.29.

$$\hat{\mathbf{x}}_k = \hat{\mathbf{x}}_k^- + \mathbf{K_{Kal}}(\mathbf{z}_k - \mathbf{H_k} \cdot \hat{\mathbf{x}}_k^-) \tag{2.29}$$

Člen v oklepaju enačbe 2.29 predstavlja merilno inovacijo ali residuum, ki zmanjšuje napako med prediktivno meritvijo $\mathbf{H_k} \cdot \hat{\mathbf{x}}_k^-$ in dejansko meritvijo. Ko doseže vrednost nič, se prediktivna vrednost dejansko ujema z meritvijo.

Matriko $\mathbf{K_{Kal}}$, dimenzij n $\times$ m imenujemo tudi ojačenje, ki minimizira a posteriori kovariančno napako. Minimizacijo izvršimo tako, da vstavimo enačbo 2.29 v enačbo za napako (enačba 2.27) in nato zamenjamo ustrezno veličino v enačbi 2.28. Dobimo enačbo 2.30:

$$\mathbf{P_k} = (\mathbf{I} - \mathbf{K_{Kal}}\mathbf{H_k})\mathbf{P_k}^-(\mathbf{I} - \mathbf{K_{Kal}}\mathbf{H_k})^T + \mathbf{K_{Kal}}\mathsf{R_{vk}}\mathbf{K_{Kal}}^T \tag{2.30}$$

Dobljeno enačbo 2.30 odvajamo na $\mathbf{K_{Kal}}$, izenačimo z 0 in rešimo za $\mathbf{K_{Kal}}$. Tako dobimo rekurzivno enačbo za *Kalmanovo ojačanje* $\mathbf{K_{Kal}}$:

$$\mathbf{K_{Kal}} = \mathbf{P_k}^-\mathbf{H_k}^T(\mathbf{H_k}\mathbf{P_k}^-\mathbf{H_k}^T + \mathsf{R_{vk}})^{-1} \tag{2.31}$$

V zgornji enačbi lahko opazimo, če limitira merilna kovariančna matrika $\mathsf{R_{vk}}$ proti vrednosti nič, potem ojačenje konvergira proti residuumu.

$$lim_{R_{vk} \to 0}\mathbf{K_{Kal}} = \mathbf{H_k}^{-1} \tag{2.32}$$

in če se a priori kovariančna napaka približuje nič, residuum izgublja pomen:

$$lim_{P_k^- \to 0}\mathbf{K_{Kal}} = 0 \tag{2.33}$$

Torej lahko potegnemo preprost zaključek glede Kalmanovega ojačenja. Če je merilna kovariančna matrika $\mathsf{R_{vk}}$ blizu vrednosti nič, potem je zaupanje v dejansko merjeno vrednost večje in v ocenjeno $\mathbf{H_k} \cdot \hat{\mathbf{x}}_k^-$ manjše. Obratno pa velja, če se a priori ocenjena kovariančna napaka približuje vrednosti nič, saj s tem dobi prediktivna vrednost $\mathbf{H_k} \cdot \hat{\mathbf{x}}_k^-$ večji pomen [73].

Delovanje rekurzivnega Kalmanovega filtra lahko povzamemo in opišemo v zanki. Iz začetnih pogojev $\hat{x}(0)^-$ in $\mathbf{P_0}^-$ izračunamo ojačanje $\mathbf{K_{Kal}}$:

$$\mathbf{K_{Kal}} = \mathbf{P_k}^-\mathbf{H_k}^T(\mathbf{H_k}\mathbf{P_k}^-\mathbf{H_k}^T + \mathsf{R_{vk}})^{-1} \tag{2.34}$$

V naslednjem koraku popravimo oceno z meritvijo:

$$\hat{\mathbf{x}}_k = \hat{\mathbf{x}}_k^- + \mathbf{K_{Kal}}(\mathbf{z}_k - \mathbf{H_k} \cdot \hat{\mathbf{x}}_k^-) \tag{2.35}$$

Popravljena ocena stanj $\hat{\mathbf{x}}_k$ je hkrati izhod filtra. Rekurzivni postopek izračunavanja nadaljujemo z izračunom kovariančne napake popravljene ocene stanj:

$$\mathbf{P_k} = (\mathbf{I} - \mathbf{K_{Kal}} \cdot \mathbf{H_k})\mathbf{P_k^-} \tag{2.36}$$

Potem z modelom napovemo prihodnje stanje sistema:

$$\begin{aligned} \hat{\mathbf{x}}_{k+1}^- &= \phi_k \hat{\mathbf{x}}_k + \mathbf{B}\mathbf{u}_k \\ \mathbf{P}_{k+1}^- &= \phi_k \mathbf{P}_k \phi_k^T + \mathsf{Q}_{\mathbf{w}k} \end{aligned} \tag{2.37}$$

Rekurzivni postopek izračuna Kalmanovega filtra se zaključi z enačbo 2.37. Postopek se nato ciklično ponavlja v zanki.

2.5.2 Določanje kota golena

Goniometri, nameščeni v kolenskem in gleženjskem sklepu, vnašajo v sistem precej negotovosti zaradi namestitve na kožo. Posledica negotovosti je zmanjšana ponovljivost meritve, kar v sistemu, ki temelji na ponavljajočem učenju hoje, ni sprejemljivo. V merilnem sistemu so na voljo pospeškometri in žiroskop, senzorji, ki ob uporabi ustreznega algoritma nudijo želeno informacijo. S kotom θ med golenom in vertikalo lahko popolnoma nadomestimo kolenski goniometer, ki je vir nezanesljivega proženja električne stimulacije.

Določanje kota golena temelji na združevanju (integraciji) dveh senzornih skupin, žiroskopa in pospeškometrov. Vsaka posamezna skupina senzorjev ima pomanjkljivosti, ki jih druga nima. S pospeškometri lahko določimo kot golena v statičnih razmerah, tj v nizkofrekvenčnem področju, kjer je absolutna vrednost dinamične komponente pospeška proti gravitacijskemu pospešku zanemarljiva. V dinamičnih razmerah, predvsem pri dotiku pete s tlemi, omenjena komponenta ni zanemarljiva. Kot golena določimo z integracijo signala žiroskopa. Signal žiroskopa vsebuje enosmerno komponento, lezenje, ki z integracijo postane rampa, zato integrirana informacija ni zanesljiva v nizkofrekvenčnem področju. Z uporabo Kalmanovega filtra, ki na podlagi modela žiroskopa napoveduje prihodnje stanje in ga popravlja z meritvijo iz pospeškometrov, dosežemo optimalno določanje kota golena v celotnem frekvenčnem področju.

Žiroskop vsebuje nizkofrekvenčno šumno komponento, t.i. lezenje. Take motnje ne moremo smatrati kot belega šuma in ga odpraviti s standardnim filtriranjem. Rešitev je v modeliranju lezenja in vključitvi v Kalmanov filter.

Model žiroskopa lahko zapišemo:

$$\begin{aligned} \dot{\theta} &= \omega + b + n_r \\ \dot{b} &= n_\omega \end{aligned} \tag{2.38}$$

oz.

$$\begin{bmatrix} \dot{\theta} \\ \dot{b} \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \theta \\ b \end{bmatrix} + \begin{bmatrix} \omega \\ 0 \end{bmatrix} + \begin{bmatrix} n_r \\ n_\omega \end{bmatrix} \tag{2.39}$$

kjer kotno hitrost sestavljajo b lezenje, ω meritev žiroskopa in nekorelirana gaussova šuma z lastnostmi:

$$\begin{aligned} E[\mathbf{n}_r] &= 0 \\ E[\mathbf{n}_r \mathbf{n}_r^{'}] &= N_r \delta(t - t^{'}) \\ E[\mathbf{n}_\omega] &= 0 \\ E[\mathbf{n}_\omega \mathbf{n}_\omega^{'}] &= N_\omega \delta(t - t^{'}) \end{aligned} \tag{2.40}$$

Zapišemo idealni enačbi:

$$\begin{aligned} \dot{\theta}_i &= \omega + b_i \\ \dot{b}_i &= 0 \end{aligned} \tag{2.41}$$

Od enačb 2.38 odštejemo idealni enačbi in tako dobimo enačbe stanj napak:

$$\begin{bmatrix} \dot{\theta} - \dot{\theta}_i \\ \dot{b} - \dot{b}_i \end{bmatrix} = \begin{bmatrix} b - b_i \\ 0 \end{bmatrix} + \begin{bmatrix} n_r \\ n_\omega \end{bmatrix} \tag{2.42}$$

ali

$$\begin{bmatrix} \Delta\dot{\theta} \\ \Delta\dot{b} \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \Delta\theta \\ \Delta b \end{bmatrix} + \begin{bmatrix} n_r \\ n_\omega \end{bmatrix} \tag{2.43}$$

Zanima nas kot θ, ki ga želimo oceniti iz žiroskopa, zato zapišemo enačbo za napako kota:

$$\Delta z = \begin{bmatrix} 1 & 0 \end{bmatrix} \begin{bmatrix} \Delta\theta \\ \Delta b \end{bmatrix} + n_\theta \tag{2.44}$$

Enačbi 2.43 in 2.44 lahko ob upoštevanju:

$$\mathbf{F} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \quad \mathbf{H} = \begin{bmatrix} 1 & 0 \end{bmatrix} \tag{2.45}$$

zapišemo v prostoru stanj:

$$\begin{aligned} \mathbf{\Delta\dot{x}} &= \mathbf{F\Delta x} + \mathbf{n} \\ \Delta z &= \mathbf{H\Delta x} + n_\theta \end{aligned} \tag{2.46}$$

Matrike merilnega šuma $\mathsf{R}_{\mathbf{v}k}$ in šuma sistema $\mathsf{Q}_{\mathbf{w}k}$ zapišemo:

$$\begin{aligned} \mathsf{Q}_{\mathbf{w}k} &= \begin{bmatrix} N_r & 0 \\ 0 & N_\omega \end{bmatrix} \\ \mathsf{R}_{\mathbf{v}k} &= N_\theta \end{aligned} \tag{2.47}$$

Dobljene enačbe uporabimo kot osnovne za izračun Kalmanovega filtra (enačbe od 2.34 do 2.37). Po rekurzivni enačbi [74] izračunamo kovariančno matriko $\mathbf{P}$:

$$\dot{\mathbf{P}} = \mathbf{FP} + \mathbf{PF}^T + \mathsf{Q}_{\mathbf{w}k} - \mathbf{P} - \mathbf{PH}^T\mathsf{R}_{\mathbf{v}k}^{-1}\mathbf{HP} \tag{2.48}$$

po enačbi 2.31 pa izračunamo Kalmanovo ojačanje:

$$\mathbf{K_{Kal}} = \begin{bmatrix} \sqrt{\frac{N_r + 2\sqrt{N_\omega N_\theta}}{N_\theta}} \\ \sqrt{\frac{N_\omega}{N_\theta}} \end{bmatrix} = \begin{bmatrix} k_1 \\ k_2 \end{bmatrix} \tag{2.49}$$

Enačba Kalmanovega filtra (po sliki 2.11) ($\Delta z = \Delta\theta$) je sedaj:

$$\frac{d}{dt}\begin{bmatrix} \Delta\hat{\theta} \\ \Delta\hat{b} \end{bmatrix} = \begin{bmatrix} 0 & 1 \\ 0 & 0 \end{bmatrix} \begin{bmatrix} \Delta\hat{\theta} \\ \Delta\hat{b} \end{bmatrix} + \begin{bmatrix} k_1 \\ k_2 \end{bmatrix} (\Delta z - \Delta\hat{\theta}) \tag{2.50}$$

kjer sta k_1 in k_2 Kalmanovi ojačanji. Če upoštevamo, da v diferenčni enačbi nastopa razlika med ocenjeno in merjeno vrednostjo, lahko diferenčne vrednosti v enačbi 2.50 nadomestimo z razliko:

$$\begin{aligned} \Delta\hat{\theta} &= \hat{\theta} - \theta \\ \Delta\hat{b} &= \hat{b} - b \end{aligned} \tag{2.51}$$

in izrazimo ocenjeno vrednost $\hat{\theta}$:

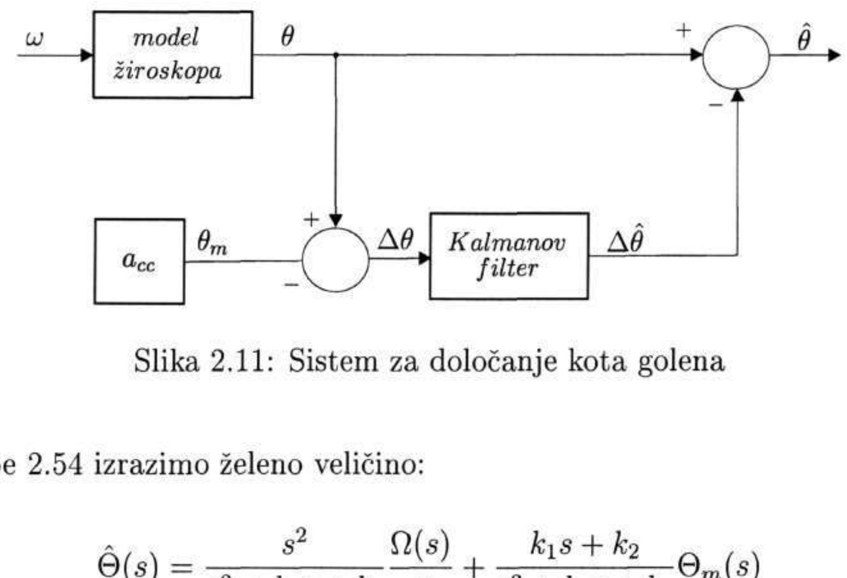
$$\frac{d}{dt}\begin{bmatrix}\hat{\theta}\\ \hat{b}\end{bmatrix} = \begin{bmatrix}0 & 1\\ 0 & 0\end{bmatrix}\begin{bmatrix}\hat{\theta}\\ \hat{b}\end{bmatrix} + \begin{bmatrix}1\\ 0\end{bmatrix}\omega + \begin{bmatrix}k_1\\ k_2\end{bmatrix}(\theta_m - \hat{\theta}) \tag{2.52}$$

kjer je θ_m merjena vrednost kota, izračunana iz pospeškometrov po enačbi:

$$\theta_m[^o] = [(\arctan\frac{a_{t0}}{a_{r0}} - \frac{\pi}{2})]\cdot(-\frac{\pi}{180}) \tag{2.53}$$

Enačbo 2.52 najenostavneje analiziramo v frekvenčnem prostoru, zato izvedemo Laplaceovo transformacijo:

$$s\begin{bmatrix}\hat{\Theta}(s)\\ \hat{B}(s)\end{bmatrix} = \begin{bmatrix}0 & 1\\ 0 & 0\end{bmatrix}\begin{bmatrix}\hat{\Theta}(s)\\ \hat{B}(s)\end{bmatrix} + \begin{bmatrix}1\\ 0\end{bmatrix}\Omega(s) + \begin{bmatrix}k_1\\ k_2\end{bmatrix}(\Theta_m(s) - \hat{\Theta}(s)) \tag{2.54}$$



Slika 2.11: Sistem za določanje kota golena

Iz enačbe 2.54 izrazimo želeno veličino:

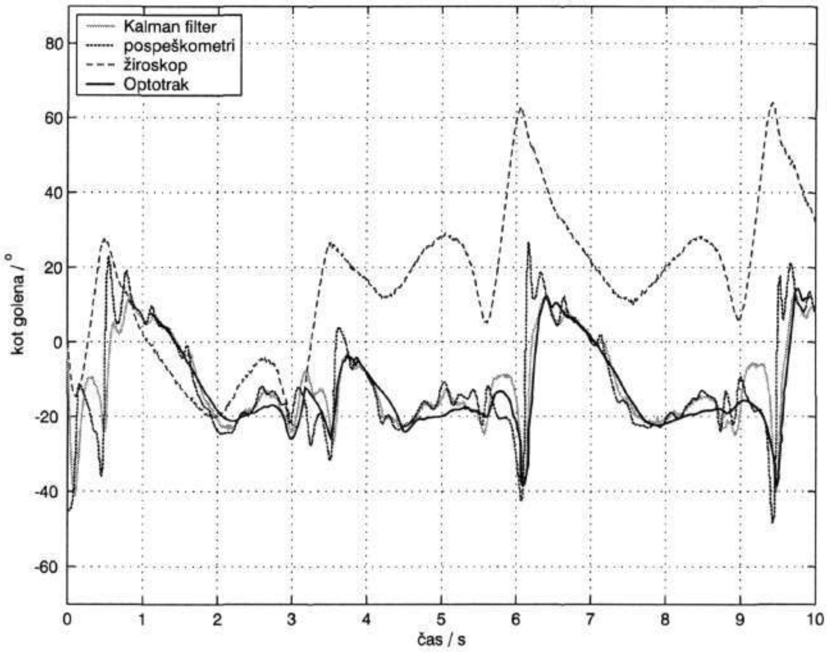
$$\hat{\Theta}(s) = \frac{s^2}{s^2 + k_1 s + k_2}\frac{\Omega(s)}{s} + \frac{k_1 s + k_2}{s^2 + k_1 s + k_2}\Theta_m(s) \tag{2.55}$$

ali v preprostejši obliki:

$$\hat{\Theta}(s) = \mathbf{G}(s)\frac{\Omega(s)}{s} + (1 - \mathbf{G}(s))\Theta_m(s) \tag{2.56}$$

Slednja enačba nam nazorno kaže, da indirektni Kalmanov filter obteži dva različna vira informacij (integriran signal žiroskopa in absolutno merjeno orientacijo), kar se odraža v komplementarnem delovanju. Funkcija $\mathbf{G}(s)$, ki filtrira integriran signal žiroskopa, deluje kot visokoprepustni filter, kar pomeni, da je pri visokih frekvencah oziroma hitrih gibih izhod enak integriranemu signalu žiroskopa. Funkcija $1 - \mathbf{G}(s)$

pa filtrira absolutni vir kota in se obnaša kot nizkoprepustni filter. Tako pri počasnih gibih izhod filtra daje prednost absolutnemu viru kota.



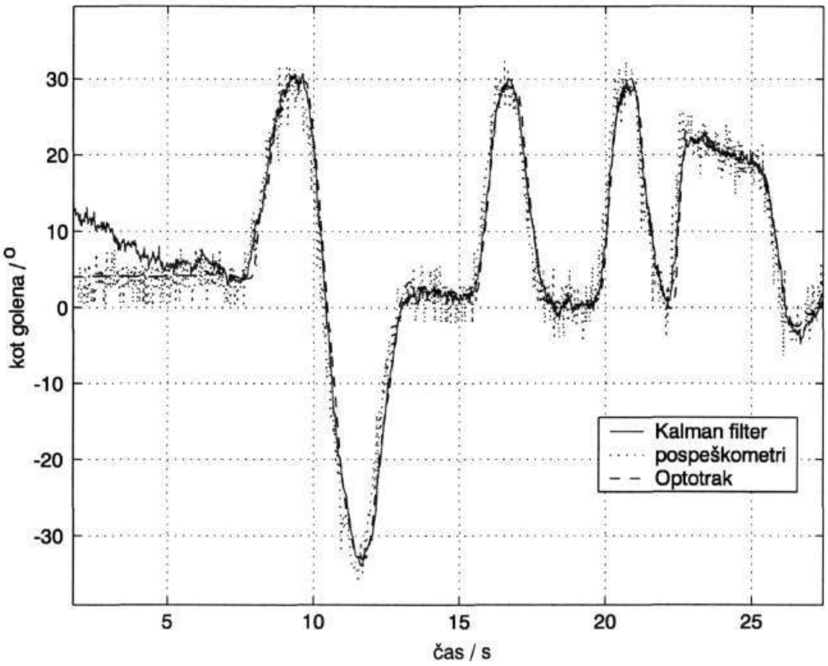
Slika 2.12: Primerjava meritev pri določanju kota golena s pospeškometri, žiroskopom, Kalmanovim filtrom in Optotrakom® pri hoji pacienta LG po ravnem terenu.

Izračunani Kalmanov filter se nanaša na shemo na sliki 2.11. Ocenjuje napako kota, ki se spreminja v skladu s karakteristikami filtra glede na določeno vrednost kota iz pospeškometrov.

Slika (2.12) prikazuje napako kota golena pri integriranju signala žiroskopa. Dodana sta tudi izračunan kot golena iz pospeškometrov po enačbi 2.53 in popravljena vrednost integriranega signala žiroskopa s Kalmanovim filtrom. Natančno delovanje smo preverili še z optičnim merilnim sistemom. Testirali smo tudi novejšo različico strojne opreme (pospeškometri ADXL202/210, žiroskop Murata ENC05 z mikrokrmilniškim nadzorom) in sensor pritrdili na goleno, na tri ustrezne točke pa pritrdili markerje optičnega merilnega sistema Optotrak®. (slika 2.13).

2.5.3 Določanje relativnega položaja spodnje ekstremitete

Tako pri analizi hoje, kakor pri ponovnem učenju hoje, se lahko poslužimo položaja ekstremitete. V začetnem poglavju o kinematiki smo za določanje absolutnega položaja uporabili optični merilni sistem *Optotrak®*. Z uporabo večsenzorne naprave, ki jo sestavljajo pospeškometri in žiroskop, lahko po vzoru inercialnih navigacijskih sistemov



Slika 2.13: Primerjava meritev pri določanju kota golena s pospeškometri, Kalmanovim filtrom in Optotrakom® z novo različico senzorja.

(INS) [75, 71, 76] določimo relativni položaj spodnje ekstremitete glede na izbrano začetno lego. Pojavi se problem lezenja signalov in šum pospeškometrov. Slednjega lahko zadovoljivo odstranimo s filtriranjem, lezenje pa ocenimo z estimatorjem.

Uporabili smo večsenzorno napravo, oba para pospeškometrov, s katerima smo določili pospešek v gleženjskem sklepu (poglavje 2.1) [22]. Za ta sklep lahko zapišemo enačbe gibanja v lokalnem, koordinatnem sistemu gležnja:

$$\begin{aligned} \ddot{x}_t &= a_{t0} - \omega \cdot \dot{z}_r - g \cdot sin\theta \\ \ddot{z}_r &= a_{r0} + \omega \cdot \dot{x}_t + g \cdot cos\theta, \end{aligned} \tag{2.57}$$

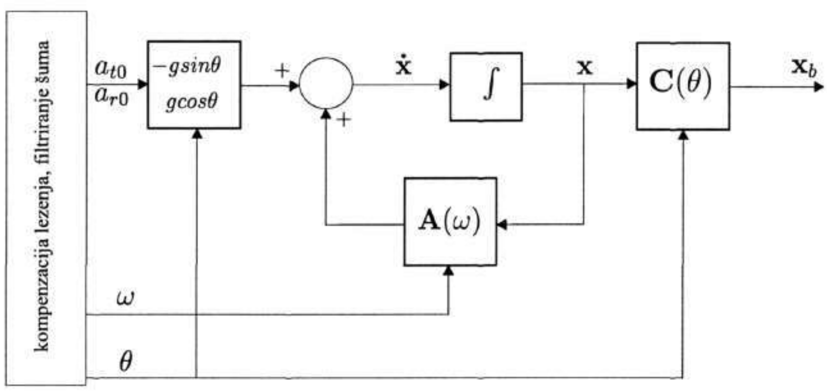
kjer sta a_{t0} in a_{r0} merjena pospeška v gleženjskem sklepu, člena $\omega \cdot \dot{z}_r$ in $\omega \cdot \dot{x}_t$ pa Coriolisova pospeška. Z zadnjima členoma enačbe odstranimo vpliv gravitacije, ki ga senzorji prav tako zajamejo v svoji meritvi. Izhodna enačba vsebuje transformacijsko matriko $\mathbf{C}(\theta)$, ki določa izhodni spremenljivki x_b in z_b. Kot θ smo določili s Kalmanovim filtrom (poglavje 2.5.2). Enačbe preoblikujemo v prostor stanj in shemo realiziramo v programskem paketu *Simulink/Matlab*®:

$$
\begin{aligned}
\ddot{x}_1 &= x_2 \\
\ddot{x}_2 &= \ddot{x}_t \\
\ddot{z}_1 &= z_2 \\
\ddot{z}_2 &= \ddot{z}_r
\end{aligned}
\tag{2.58}
$$

$$
\begin{bmatrix} \dot{x}_1 \\ \dot{x}_2 \\ \dot{z}_1 \\ \dot{z}_2 \end{bmatrix} = \begin{bmatrix} 0 & 1 & 0 & 0 \\ 0 & 0 & 0 & -\omega \\ 0 & 0 & 0 & 1 \\ 0 & \omega & 0 & 0 \end{bmatrix} \begin{bmatrix} x_1 \\ x_2 \\ z_1 \\ z_2 \end{bmatrix} + \begin{bmatrix} 0 \\ a_{t0} - g \cdot sin\theta \\ 0 \\ a_{r0} + g \cdot cos\theta \end{bmatrix} \tag{2.59}
$$

$$
\begin{bmatrix} x_b \\ z_b \end{bmatrix} = \begin{bmatrix} -cos\theta & 0 - sin\theta & 0 & \\ sin\theta & 0 & -cos\theta & 0 \end{bmatrix} \begin{bmatrix} x_1 \\ x_2 \\ z_1 \\ z_2 \end{bmatrix} \tag{2.60}
$$

Zapis v prostoru stanj sicer ne zmanjša števila integratorjev, a s tem odpade potreba po dvakratnem integriranju signala. Integrator ojači vsako najmanjšo napako. Tako napaka zaradi enosmerne komponente, ki je lahko rezultat neprecizne kompenzacije lezenja, z integracijo narašča s časom.



Slika 2.14: Sistem za določanje relativnega položaja sklepa

Kljub precej dobri in razmeroma učinkoviti ideji se je izkazalo, da je končna napaka prevelika za potrebe ponovnega učenja hoje. Pri ponovnem učenju hoje namreč zahtevamo ponovljivost merjene veličine, saj informacijo uporabljamo za vodenje električne stimulacije. Za ponovno učenje hoje pa moramo zagotoviti ponovljivost postopka [49]. Ker ponovljivosti ne moremo zagotoviti, smo se odločili, da za vodenje električne stimulacije uporabimo kot golena, kjer je ponovljivost zagotovljena.

3

Kognitivna povratna zanka kot vir informacije

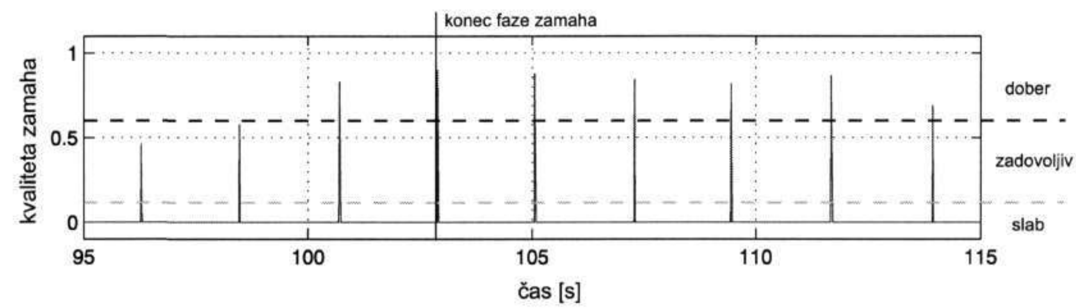
Senzorne informacije v sistemu za ponovno učenje hoje ne uporabimo samo za analizo, pač pa tudi za zaprtozančno vodenje. Pri zaprtozančnem vodenju imamo velikokrat v mislih regulator, ki regulira izhodno veličino na podlagi meritev, senzorne informacije. O zaprtozančnem vodenju lahko govorimo tudi takrat, ko je v zanko vključen človek, ki ustrezno glede na sprejeti signal vodi regulirno veličino in s tem vpliva na izhodni signal [77]. V tem primeru je potrebno signal, ki je namenjen človeku, oblikovati tako, da ga je sposoben sprejemati, razločiti, hkrati pa tako opravilo ne sme zmotiti njegove aktivnosti pri opravljanju zadane naloge [62].

Signali, zajeti neposredno s senzorji ali že izračunane veličine iz poglavja 2, so zvezne veličine, ki predstavljajo preobsežno informacijo. Take informacije človek ni sposoben zaznavati in se ustrezno odzvati med opravljanjem druge dejavnosti, pri kateri mora biti osredotočen še na mnoge druge dejavnike. Kadar imamo v mislih hojo, nastopi poleg vseh ostalih dejavnikov kot so hitrost hoje, ovire, še kinematična in dinamična stabilnost [2]. Tako zahtevajo omenjeni dejavniki precejšnjo pozornost in s tem zmanjšujejo možnost dojemanja senzorne informacije, kar je še posebej opazno pri osebah z gibalnimi motnjami, hemiplegijo, poškodbo hrbtenjače, multiplo sklerozo. V ta namen je bilo potrebno skrbno izbrati informacijo, ki je namenjena osebi in jo poenostaviti. Možna in hkrati precej dobra izbira pri poenostavitvi je diskretizacija senzorne informacije.

Razdelitev hoje v fazo zamaha in fazo opore nam že sama ponuja informacijo, ki jo lahko posredujemo osebi. Temu je bil dodan še algoritem za določanje kvalitete zamaha, kar se je kasneje izkazalo [78] kot dobra izbira. Izhod razvitega algoritma (poglavje 2.4) je zvezen, zato je bil diskretiziran. Pri diskretizaciji sem upošteval gibalne sposobnosti osebe, a ker je referenčni zamah zajet pri vsaki osebi posebej, torej relativno, ni potrebe po vsakokratnem nastavljanju pragov diskretnih signalov. Zaradi omenjene kompleksnosti in sposobnosti percepcije človeka med hojo so bili izbrani trije diskretni nivoji, ki na podlagi ocene kvalitete zamaha (enačba 2.22) podajajo informacijo o zamahu kot *'slab'*, *'zadovoljiv'* in *'dober'* (slika 3.1).

`če kvaliteta zamaha` > 0 in < 0.1 --→ zamah *'slab'*
`če kvaliteta zamaha` > 0.1 in < 0.6 --→ zamah *'zadovoljiv'*
`če kvaliteta zamaha` > 0.6 --→ zamah *'dober'*

Upoštevamo, da vrednost 1 predstavlja popolno ujemanje izmerjenega z referenčnim zamahom.



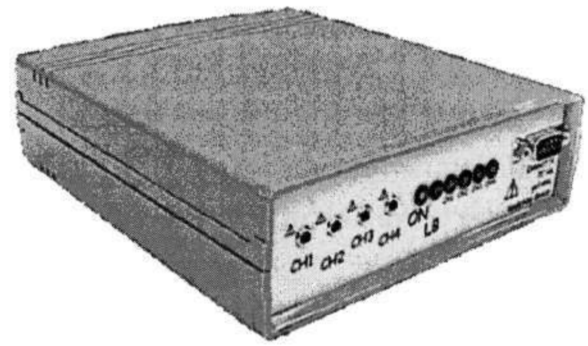
Slika 3.1: Kognitivna povratna zanka nudi diskretno informacijo (slab, zadovoljiv, dober) o kvaliteti zamaha. Nivoji med posameznimi ocenami so nastavljivi.

Po diskretizaciji informacije, ki jo želimo posredovati osebi, je bilo potrebno izbrati najustreznejši način posredovanja povratne informacije. Glede na izbrano strategijo rehabilitacije, kjer oseba hodi po traku za hojo, je potrebno zagotoviti čim manj strojne opreme, ki bi osebo ovirala pri hoji. Boljše možnosti ponuja uporaba kognitivne povratne zanke, pri kateri lahko vsako diskretno stanje predstavimo kot zvočni pisk različnih frekvenc ali zvokov. Izkazalo se je, da je oseba pri hoji na traku (0.7 km/h) lahko razpoznala tri tone, ki predstavljajo *'slab'*, *'zadovoljiv'* ali *'dober'* zamah [79]. Hkrati je fizioterapevtka, ki je prav tako spremljala zvočni signal, imela možnost verbalno poučiti osebo o kvaliteti pravkar izvedenega zamaha. Uporaba zvočne kognitivne povratne zanke kot vira informacije o fazi zamaha razbremenjuje fizioterapevte in osebo, ki se lahko osredotoči na najpomembnejšo nalogo, hojo po traku. Hkrati uporaba kognitivne povratne zanke povzroči visoko stopnjo motivacije pri pacientu.

4

FES kot motorični pripomoček pri hoji paretičnih oseb

V uvodu je bila na kratko predstavljena funkcionalna električna stimulacija kot terapevtsko in funkcionalno orodje za doseganje funkcionalnih gibov, ki so posledica krčenja posameznih mišic, kar dosežemo z električnimi impulzi. Funkcionalni gib ekstremitete lahko dosežemo z eferentno ali aferentno FES. Pri ljudeh s poškodbo hrbtenjače so povezave med osrednjim in perifernim živčevjem popolnoma ali delno prekinjene, odvisno ali gre za plegijo ali parezo. V takih primerih poskušamo naravni motorični signal nadomestiti z električno stimulacijo. S stimulacijo eferentnih poti doseženo neposredni gib, medtem ko s stimulacijo aferentnih (senzoričnih) poti sprožimo refleksni gib, kjer spinalni center sproži impulz motoričnega nevrona. Pri pacientih z nepopolno poškodbo hrbtenjače, ki imajo ohranjene nekatere senzorne in gibalne sposobnosti, lahko s pridom izkoristimo fleksijski refleks, tj hkratno fleksijo kolka in kolena, ter dorsifleksijo gležnja. Pri tem uporabimo enokanalno stimulacijo peronealnega živca [34]. Tukaj se ponuja kar nekaj možnosti glede uporabe stimulatorjev [80]. Peronealni stimulatorji so načeloma lahko majhni implantibilni [81], a ker gre v večini primerov za terapijo ali začasno uporabo, je smiselno uporabiti površinske zunanje stimulatorje [2, 82, 83]. V predlaganem rehabilitacijskem postopku je uporabljen programabilen električni stimulator, razvit v Laboratoriju za biomedicinsko tehniko in robotiko [84] (slika 4.1), saj strategija zahteva možnost reguliranja jakosti stimulacije v realnem času.



Slika 4.1: 4 kanalni programabilni električni stimulator, razvit v Laboratoriju za biomedicinsko tehniko in robotiko [84].

4.1 Elektrode pri FES

Stimulacija nevronov poteka preko para elektrod, ki predstavljata neposreden stik s pacientom. Električni tok se na elektrodah pretvori v tok ionov, obe elektrodi ustvarita električno polje, v katerem je mišično-živčno tkivo dielektrik. Odvisno od vrste elektrod in uporabe napetostnega ali tokovnega el. stimulatorja, pri površinski stimulaciji lahko ob nepravilni namestitvi povzročimo poškodbe kože. Od načina pritrditve je odvisno električno polje in s tem tudi ponovljivost in učinkovitost s stimulacijo doseženih gibov.

Poleg delitve na površinske (nameščene na koži) in implantirane (ob živcu in v mišici) elektrode ni nepomembna delitev na unipolarne in bipolarne elektrode. Pri unipolarnih je ena elektroda manjša kakor druga in je električno polje koncentrirano okoli manjše. Pri nameščanju elektrod je predvsem pomembna razdalja med elektrodama. Če sta elektrodi blizu skupaj, se tok predvsem zaključuje po površini, kar lahko ob primerni vlažnosti kože povzroči neprijeten pekoč občutek, medtem ko z večjo razdaljo dosegamo globja tkiva [2]. Vendar je s tem večja tudi upornost in poraba energije. Druga nevšečnost, ki se pogosto pojavlja, je položaj površinskih elektrod. Vsakokratno nameščanje povzroči neenak položaj, kar se odraža v spremenjenem odzivu mišice in posledično gibu okončine.

Vsem tem nevšečnostim se izognemo z uporabo implantiranih elektrod, ki jih namestimo vzdolž živca ali okrog živca (cuff electrodes) [11, 8]. S tem zmanjšamo tudi porabo energije, kar je bistvenega pomena. V tem primeru je nujen kirurški poseg, elektrodam zraste cena in v primeru, ko ne gre za permanentno uporabo pač pa za terapevtski rehabilitacijski postopek, to ni smiseln pristop. Razvoj perkutanih elektrod [85], ki jih z iglo namestimo v mišico, ima nekatere prednosti implantibilnih elektrod in nekatere prednosti površinskih elektrod. Ker so nameščene v notranjosti mišice, je povečana selektivnost električne stimulacije. Namestitev z iglo in zunanja povezava z žico pa izloči potrebo po operativnem posegu.

V raziskavi smo v kombinaciji z omenjenim 4-kanalnim električnim stimulatorjem uporabili gumbaste površinske elektrode ali novejše tanke lepljive (Axelgaard Manufacturing Co., Ltd.). elektrode za enokanalno stimulacijo peronealnega živca, s katero smo izzvali fleksijski odziv. Le v izrednem primeru smo uporabili dodatni stimulacijski kanal, s katerim smo reševali ostale težave pri gibanju spodnjih okončin, npr. nesposobnost fleksije kolenskega sklepa.

4.2 Način vodenja stimulacije

Največjo težavo pri električni stimulaciji predstavlja vodenje. Ker je hoten gib zaradi poškodbe osrednje živčne poti onemogočen, so za funkcionalen gib potrebni električni impulzi, ki jih nudi električni stimulator. Ob tem se pojavi problem, kdaj sprožiti tak vlak impulzov, kako določiti njegovo trajanje in kako ga sprožiti. Za proženje lahko uporabimo druge mišice, kjer merimo elektromiogramske (EMG) signale, direktni signal motoričnih živcev z merjenjem elektronevrograma (ENG) ali se poslužimo drugih zunanjih možnosti, npr. petnega stikala [34], ročne tipke v bergli [2], morda celo ročice [77], s katero imamo možnost uravnavati amplitudo stimulacije. Pri slednjem pristopu mora oseba zavestno sprožiti stimulacijo, kar pomeni, da ta naloga ne sme ovirati procesa, v katerem želi izvesti funkcionalni gib.

Omenjeni 4-kanalni stimulator je programsko krmiljen preko RS232 vmesnika. Namenjen je uporabi skupaj z nadzornim procesorjem oz kar osebnim računalnikom. Pri inicializaciji podamo število stimulacijskih kanalov, ki jih bomo uporabljali, in frekvenco stimulacije. Med stimulacijo sproti pošiljamo podatke o širini impulza (v μs) in o amplitudi izhodnega toka (v mA). Obe veličini skupaj predstavljata naboj električne stimulacije. Proženje stimulacije je torej popolnoma prepuščeno nadzornemu računalniku. Uporabnik lahko programsko določi trenutek proženja, morebitno zakasnitev, lahko ga poveže s hotenim dogodkom:

- dodatna prožilna tipka v merilnem sistemu, vendar je v tem primeru jakost stimulacije konstantna

- prožilna ročica, kjer se spreminja jakost stimulacije.

V obeh primerih je proženje električne stimulacije ročno. Stimulacijo lahko proži uporabnik ali fizioterapevt. Tako proženje postane predvsem zahtevno pri večkanalni stimulaciji, kjer je potrebno predprogramirati vsako sekvenco električnih impulzov. S pritiskom na prožilno tipko sprožimo hkratno stimulacijo vseh kanalov, ki so ustrezno zakasnjeni. Prednost ročnega proženja je hoteno sodelovanje osebe v procesu rehabilitacije [30, 86]. Alternativna možnost, ki je predvsem zanimiva za implantirano stimulacijo, je:

- avtomatsko proženje stimulacije glede na izbrano merilno veličino
- proženje stimulacije na podlagi senzorne informacije in hkratna regulacija jakosti stimulacije.

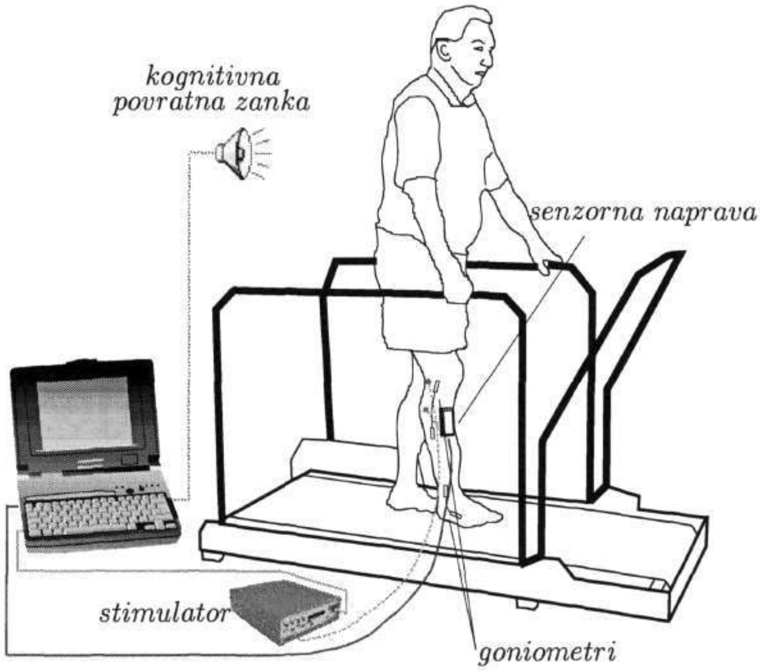
Pri prvi možnosti določimo značilko v izmerjenem signalu in ko le-ta nastopi, se sproži električna stimulacija. Tak način je lahko uporabljen pri implantirani enokanalni stimulaciji, predvsem pri peronealni stimulaciji, ko gre za problem padanja stopala, ki je največkrat posledica hemiplegije. Druga možnost predstavlja kompleksnejši problem. Kadar gre za obnovitev hoje pri paraplegikih, vstajanju para-tetraplegikov-paretikov ali pa pri kombiniranju hotenega funkcionalnega giba in električne stimulacije pri paretičnih osebah je smiselno analizirati senzorno informacijo in jo uporabiti v namene vodenje električne stimulacije.

5

Sistem za ponovno učenje hoje

Kolikor uporabimo algoritme, opisane v prejšnjih poglavjih, lahko na podlagi znanj o kinematiki spodnjih ekstremitet in poznavanju obstoječih rehabilitacijskih tehnik pri pacientih z nepopolno poškodbo hrbtenjače realiziramo idejo o nastajajočem sistemu za ponovno učenju hoje (Gait Reeducation System - GRS). Sistem temelji na merjenju in analizi kinematičnih parametrov hoje z večsenzornim sistemom in goniometri pri hoji oseb po tekočem traku. Novejšo različico večsenzornega sistema (slika 5.3) sestavljata dva dvo-osna pospeškometra (Analog Devices ADXL202/210), ki merita pospeške, kot je prikazano na sliki 2.6, in žiroskop (Murata ENC05), ki meri kotno hitrost golena. Nadzor poteka z mikrokrmilnikom Atmel, ki komunicira z osebnim računalnikom preko serijskega vmesnika. Večsenzorni sistem je z velcro trakom pritrjen na sprednjo stran golena kot prikazuje slika 5.1. Dopolnilno nalogo merjenja kinematičnih parametrov opravljata goniometra (Biometrics, Ltd.), nameščena v gleženjskem in kolenskem sklepu. Algoritem za detekcijo faze zamaha in algoritem za oceno kvalitete zamaha služita vrednotenju faze zamaha. Faza zamaha se ovrednoti po vsakem zamahu. Informacija je nato posredovana osebi v obliki zvočnega signala treh različnih frekvenc. Med hojo se shranjujejo tudi podatki o tem, koliko dobrih oz. slabih zamahov je bilo izvedenih zaporedoma.

V primeru, ko je na traku oseba z gibalnimi motnjami, v sistemu uporabimo FES sistem, ki ga predstavlja računalniško voden električni stimulator s površinskimi elektrodami. Strategija električne stimulacije temelji na oceni kvalitete faze zamaha. Pri doseganju zadovoljive kvalitete zaporednih zamahov lahko motorično podporo znižamo in obratno. Seveda pa pri tem oseba zavestno sodeluje, saj je o kvaliteti faze zamaha obveščena preko kognitivne povratne zanke in FES podpora ni nikoli tolikšna, da bi omogočala hojo brez sodelovanja osebe oz pacienta.

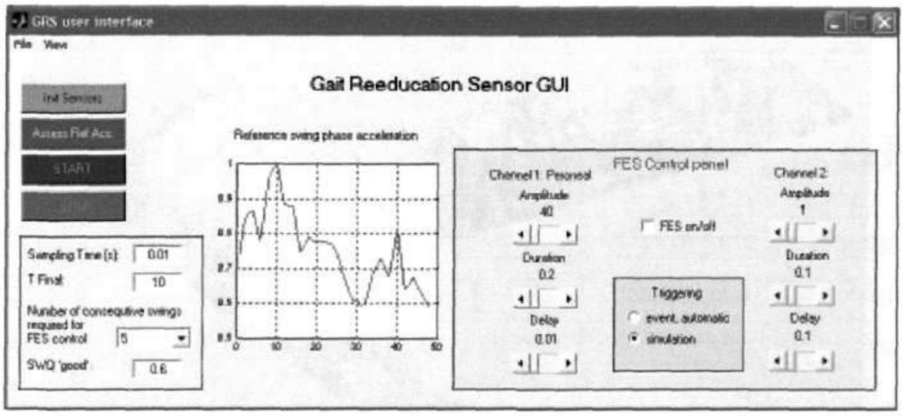


Slika 5.1: Sistem za ponovno učenje hoje sestavljajo trak za hojo, večsenzorni sistem, pritrjen na golenu z elastičnim trakom, goniometri, električni stimulator in kognitivna povratna zanka.

5.1 Ročno proženje FES pri GRS

Prva različica sistema za ponovno učenje hoje je temeljila na zasnovi, ki izhaja iz študije Erzina [86, 29]. S tem smo predvsem hoteli pacienta vključiti v rehabilitacijski sistem [30]. V večini primerov [31] so pacienti z nepopolno poškodbo hrbtenjače ali hemiplegiki uporabljali bergle kot ortotični pripomoček za hojo. Tako je bergla pripravna za namestitev tipke ali ročice za proženje električne stimulacije. Namestimo jo v ročaju bergle in pacient proži stimulacijo po potrebi. Kadar gre za enokanalno stimulacijo peronealnega živca, lahko z rehabilitacijskim treningom dosežemo že skoraj podzavestno proženje stimulacije in s tem dokaj sprejemljivo hitrost hoje. Stimulacija v tem primeru rešuje problem padanja stopala, ki zelo ovira fazo zamaha, v sistemu za ponovno učenje hoje pa z njo izzovemo fleksijski odziv.

V sistemu za ponovno učenje hoje smo predvideli uporabo traku za hojo (treadmill). Pacient je zaradi zanesljive opore rok pridobil veliko na stabilnosti, kar mu je omogočalo, da je svojo pozornost posvečal izključno hoji. Z istim namenom smo proženje električne stimulacije prepustili fizioterapevtki. S pritiskom na posebej prirejeno tipko je sprožila vlak električnih impulzov. Čas stimulacije ni bil vnaprej določen,



Slika 5.2: Uporabniški vmesnik sistema za ponovno učenje hoje. Omogoča nastavljanje parametrov stimulacije na dveh kanalih, zakasnitvenih časov, trajanja stimulacije, jakosti, števila potrebnih zamahov za oceno motorične podpore, nastavitev nivoja zahtevnosti za oceno kvalitete zamaha in zajem referenčnega pospeška.

pač pa je bil odvisen od tega, koliko časa je bila pritisnjena tipka. Parametri stimulacije so prikazani v tabeli 5.1.

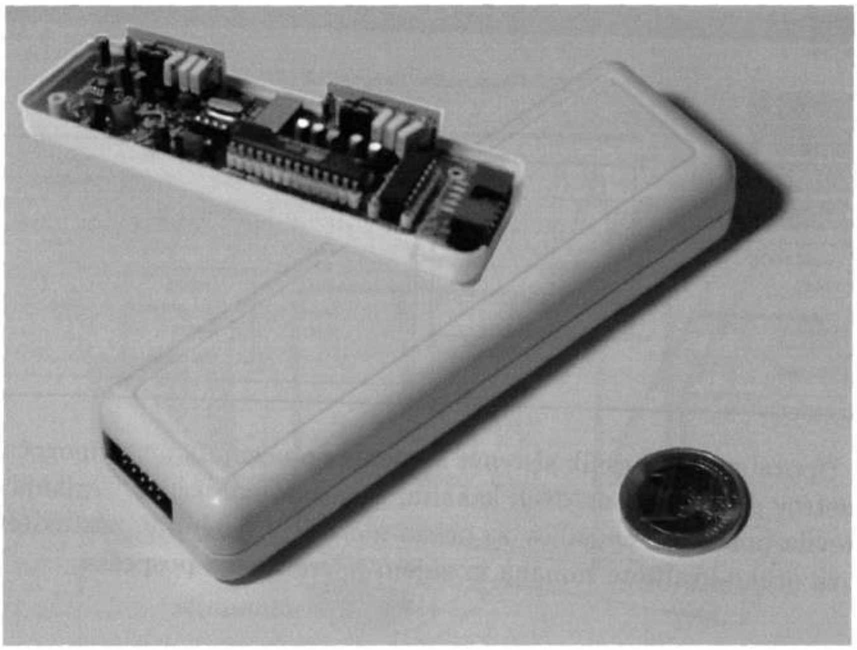
Tabela 5.1: Parametri stimulacije.

frekvenca stimulacije	širina impulza	izhodni tok	tip elektrod
20 Hz	80 μs	45 mA	površinske

5.1.1 Vrednotenje kognitivne povratne zanke

Kognitivna povratna zanka predstavlja pomemben člen pri treningu hoje na traku. Kljub vsem algoritmom, ki so opisani v poglavju 2 in s katerimi razdelimo hojo v posamezne faze ter jih ovrednotimo, je potrebno uporabniku informacijo predstaviti v dovolj enostavni obliki, da jo je sposoben sprejemati med aktivnostjo in še vseeno dovolj dobro ponazarja izračunano veličino. Faza zamaha je številčno ovrednotena in je bila diskretizirana (slika 3.1). Diskretne vrednosti, ki nudijo zadostno informacijo o fazi zamaha, so bile posredovane osebi med hojo po traku v obliki zvočnega signala. Za *'dober'* zamah je bil izbran visokofrekvenčni (3kHz, trajanja 100 ms) ton, za *'zadovoljiv'* ton s frekvenco 1 kHz, trajanja 200 ms in za *'slab'* zamah nizkofrekvečni ton (500 Hz, trajanja 200 ms). Pacienti so opisane signale lahko brez večjih težav razpoznali, manjše težave so se pojavljale pri razločevanju *'zadovoljivo'* in *'dobro'* [80].

V novejši različici sistema za ponovno učenje sem nadomestil piskajoče zvoke s



Slika 5.3: Večsenzorni sistem za merjenje kinematičnih veličin. Namestimo ga na goleno merjene osebe. Sestavljajo ga žiroskop in 2 dvoosna pospeškometra. Vse nadzoruje mikrokrmilnik Atmel^{TM}, ki komunicira z osebnim računalnikom preko RS232 vmesnika.

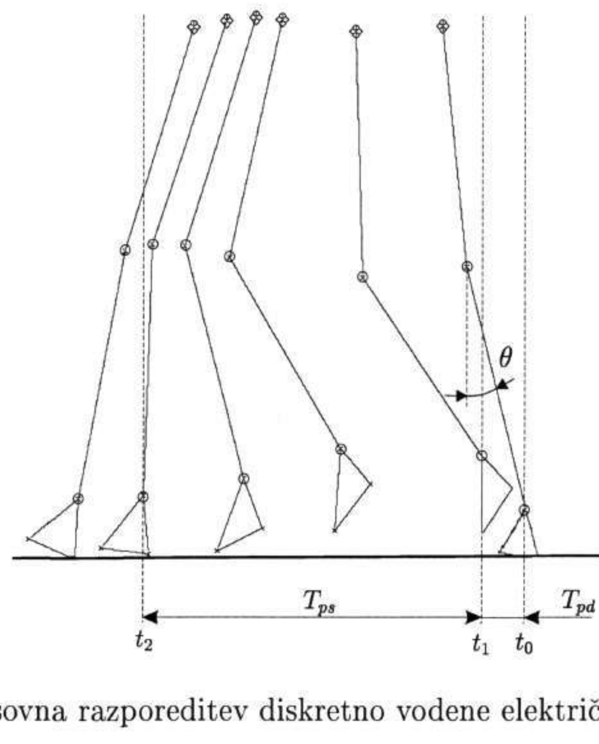
polifoničnimi zvoki, ki jih je generirala zvočna kartica. Izkazalo se je, da je tak zvok mnogo prijetnejši in hkrati tudi razpoznavnejši.

5.1.2 Uporaba dodatnega stimulacijskega kanala

Čeprav je v sistemu za ponovno učenje hoje predvidena enokanalna peronealna stimulacija, se lahko pojavi potreba po dodatnem stimulacijskem kanalu. Dodatni stimulacijski kanal pri tem služi le kot podpora peronealni stimulaciji in ne spremeni strategije oz vzorca stimulacije. Tak primer nastopi pri močno spastičnih pacientih, kjer je odziv na stimulacijo nenaden in nenadzorovan [87]. Drug primer je pomanjkanje nadzora nad nekaterimi mišicami, kar je posledica poškodbe osrednjega živčnega sistema. Posledično obstaja možnost, da pride do zaklepa sklepov, recimo, kolenskega sklepa. Tako je pacient kljub peronealni stimulaciji prikrajšan za fleksijo kolena in posledično postane njegov zamah oviran, saj je za njegovo izvedbo potrebno dvigniti kolk. Pacientu ID, ki je utrpel zoženje spinalnega kanala v vratnem predelu C2-C6, smo nudili dodatno stimulacijo fleksorjev kolena v predfazi zamaha. Kratek vlak pulzov je ob proženju peronealne stimulacije sprostil kolenski sklep in omogočil izvajanje fleksijskega refleksa.

5.2 Diskretno vodenje proženja FES pri GRS

S povečevanjem hitrosti hoje na traku postane ročno proženje električne stimulacije precej težavno. Fizioterapevtkino prizadevanje, da bi ob pravem trenutku sprožila stimulacijo, predvsem pa določila trajanje, in s tem prispevala k izboljšanju kvalitete zamaha, je bilo zaman. Možnost, da je bil trenutek proženja zgrešen ali da je bil čas stimulacije prekratek ali predolg, je lahko bistveno vplivala na kvaliteto zamaha, kljub temu da je pacient vlagal veliko truda v izvajanje 'boljšega' zamaha.



Slika 5.4: Časovna razporeditev diskretno vodene električne stimulacije.

V ta namen so bili uporabljeni algoritmi iz poglavja 2.5.2. Na podlagi kota kolena, ki ga merimo z goniometrom lahko določimo trenutek proženja stimulacije. Stimulacijo je potrebno aktivirati tik preden se začne faza zamaha. Tega z algoritmom, opisanem v poglavju 2.3.2, ni moč določiti. Tako je bil trenutek določen kar s kotom kolena, pri tem pa je bilo upoštevano, v kateri fazi hoje se oseba nahaja. Trenutek proženja smo ročno nastavili pred vsako meritvijo za vsakega pacienta posebej. Do razlike v trenutkih proženja je prišlo zaradi raznolikosti poškodb pri pacientih in posledično tudi potreb po stimulaciji. V uporabniški vmesnik (slika 5.2) smo dodali možnost nastavljanja zakasnitve T_{pd} in časa trajanja T_{ps} stimulacije ter tako zadostili zahtevam pri stimulaciji. Časovna razporeditev električne stimulacije je prikazana na sliki 5.4. Ob prednastavljenem kotu goleni (θ), ki ga sproti izračunava algoritem za določanje kota (2.5.2), se sproži električna stimulacija. Stimulacijo lahko po potrebi zakasnimo z zakasnitvijo T_{pd}.

V nadaljevanju razvoja se je izkazalo, da lahko kolenski goniometer, ki je vir nezanesljivosti, nadomestimo s kotom θ (slika 2.5.2) med golenom in vertikalno ravnino (poglavje 2.5.2). S tem postane sistem veliko robustnejši, predvsem pa privlačnejši s strani uporabnika, ki ponavadi ne mara nositi prevelike količine opreme. Tudi s praktičnega vidika veliko merilne in rehabilitacijske opreme ovira sproščeno hojo po traku.

5.3 Strategija treninga ponovnega učenja hoje

Po skrbnem načrtovanju sistema za ponovno učenje hoje je potrebno določiti strategijo učenja, ki je velikokrat odvisna od tipa poškodbe ali bolezni pacienta. Strategija, ki smo jo zasnovali, temelji na uporabi traku za hojo, ki omogoča enakomerno hojo tudi pri hujših gibalnih motnjah. Po potrebi lahko postavimo pacienta v sistem za razbremenitev teže, večina pa ohranja ravnotežje s pomočjo opornih gredi na traku. Tak pristop omogoča pacientu enakomerno hojo, pri tem pa so mu v pomoč fizioterapevti. Kombinacija s funkcionalno električno stimulacijo razbremeni delo fizioterapevta in omogoča hitrejšo hojo po traku.

Predlagan pristop, ki uporablja umetne senzorje, v določeni meri nadomesti izgubljeno senzorno informacijo, ocenjuje fazo zamaha in hkrati informacijo posreduje pacientu in fizioterapevtom. Tako dobijo oboji natančno informacijo o kvaliteti faze zamaha in lažje usmerjajo pacienta. Uporaba senzorne informacije pri rehabilitaciji s trakom za hojo in funkcionalno električno stimulacijo omogoča boljše odločanje pri proženju in trajanju motorične podpore. Pri številnih meritvah se je pokazalo, da bi lahko bil pristop z avtomatskim vodenjem električne stimulacije zelo učinkovit. Omogoča natančno definiran trenutek proženja in trajanje stimulacije, kar se odraža predvsem v ponovljivosti postopka. Novost pri takem pristopu je vodenje jakosti stimulacije glede na kvaliteto faze zamaha. Med hojo pacienta po traku izdelana senzorna naprava zajema parametre hoje, računalnik izračuna kvaliteto faze zamaha in nudi informacijo o tem v obliki zvočnega signala. Motorično podporo pri hoji pa nudimo pacientu s pomočjo funkcionalne električne stimulacije. V primeru, ko pacient med hojo uspešno opravi določeno število dobrih zaporednih zamahov, lahko motorično podporo znižamo. Posledično izzovemo večjo stopnjo sodelovanja pacienta v rehabilitacijskem postopku. Seveda velja tudi obratno, pri določenem številu zaporedno izvedenih slabih zamahov zvišamo jakost električne stimulacije.

V rehabilitacijskem postopku, ki bi zajemal dolgotrajen trening pacientov, je potrebno zastaviti strategijo. Trajanje vsakodnevne vaje je omejeno z utrujanjem pri električni stimulaciji [5], tako je trening omejen na največ eno uro. Glede na izkušnje pri klasični rehabilitaciji traja postopek trikrat tedensko od enega do treh mesecev, odvisno od pacientove zdržljivosti in tipa poškodbe.

6

Rezultati

Sam razvoj tako senzorne naprave, kakor kognitivne povratne zanke in motorične podpore, je zahteval hkratno testiranje algoritmov. Tak pristop omogoča takojšnjo odpravo napak, pomanjkljivosti, in hkratno izboljševanje celotnega sistema za ponovno učenje hoje.

Z meritvami smo pričeli v laboratoriju na ravnem terenu. Med hojo pacienta LG s pomočjo bergel smo merili dostop s pomočjo pritiskovnih plošč AMTI®, kote v gleženjskem in kolenskem sklepu z goniometri ter pospeške in kotno hitrost golena s prvo različico večsenzorne naprave [63]. Kot referenčni sistem smo uporabili optični merilni sistem Optotrak®, ki s pomočjo infrardečih kamer določa položaj markerjev, nameščenih v sklepih in na okončninah pacienta. S poznavanjem kinematike lahko določimo položaj posameznih segmentov in na podlagi trajektorij analiziramo hojo. Hkrati smo izvajali meritve na osebah brez gibalnih motenj, ki so bile podlaga za modeliranje in biomehansko analizo hoje. S senzorji izmerjene veličine so bile preračunane tako, da jih je bilo možno primerjati z meritvami, dobljenimi z optičnim merilnim sistemom, ki smo jih smatrali za referenčne.

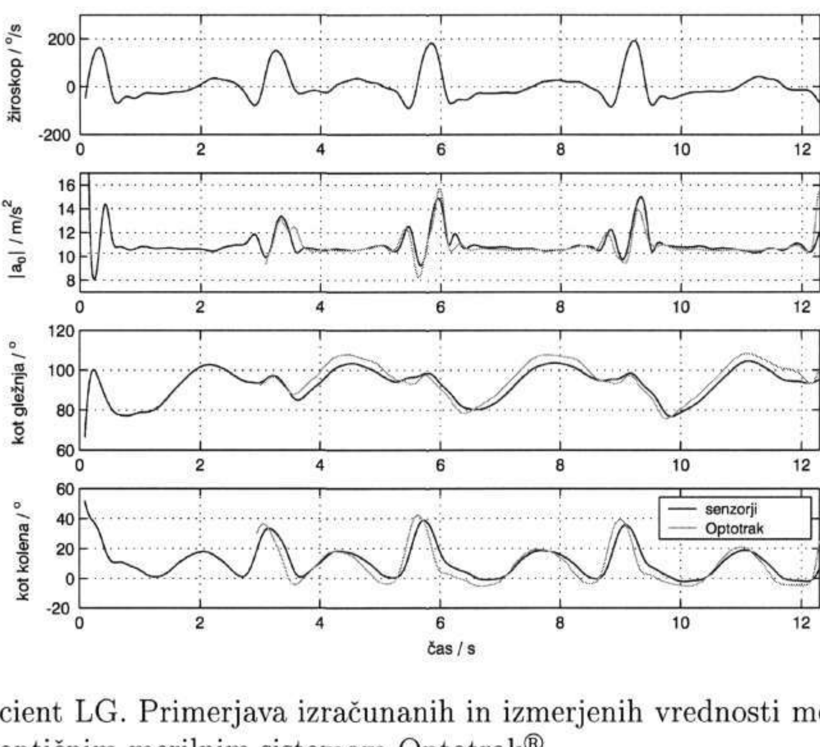
V nadaljevanju so prikazane meritve, opravljene med razvojem sistema za ponovno učenje hoje. Pri analizi videoposnetkov hoje pacienta LG po ravnem terenu je bilo opaziti, da je pacient večino pozornosti namenil vzdrževanju kinematične stabilnosti [2]. Zaradi tega bi bilo nesmiselno razvijati sistem za ponovno učenje hoje, ki bi temeljil na hoji po ravnem terenu. Hojo smo prestavili na tekoči trak z namenom, da pacientovo pozornost usmerimo na hojo in ne na vzdrževanje ravnotežja. Pacienti BM, ID in BK ter oseba ZM so sodelovali pri testiranju v različnih obdobjih nastajanja sistema.

6.1 Rezultati meritev pri pacientu LG z nepopolno poškodbo C6-7

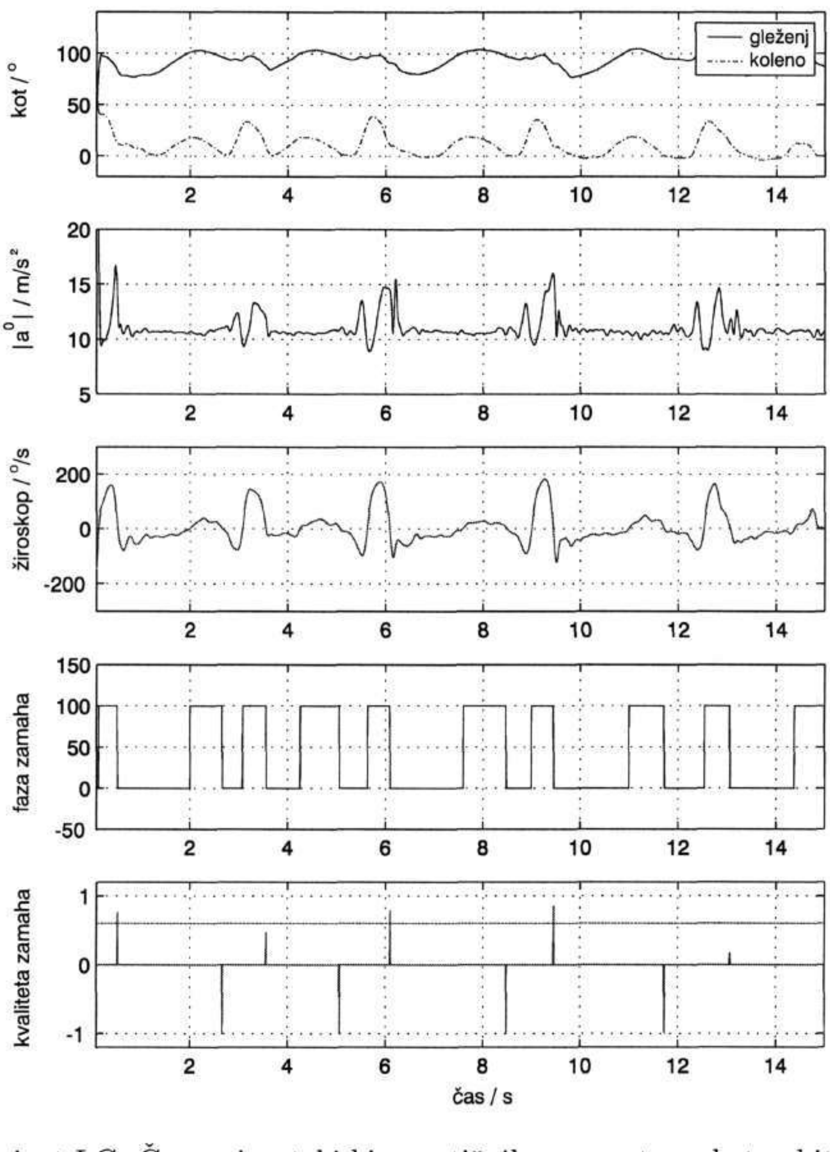
Pacient LG je hodil po ravnem terenu in pri tem uporabljal bergle. Slika 6.1 prikazuje izmerjene signale iz senzorjev v primerjavi z optičnim merilnim sistemom. Senzorna naprava, ki meri kotno hitrost golena in tangecialni ter radialni pospešek gležnja, je bila nameščena na sprednjem delu golena. S pomočjo infrardečih markerjev in optičnega sistema smo izračunali primerljive veličine. Slika 6.2 prikazuje implementacijo prvih algoritmov za detekcijo faze zamaha. Pozneje smo izmerjene podatke uporabili tudi v algoritmu za ocenjevanje kvalitete zamaha. Opazimo lahko, da so napačno detektirani zamahi obravnavani kot neustrezni in posledično je vrednost takega zamaha negativna (-1).

Tabela 6.1: Podatki pacienta LG.

Teža	Višina	Starost	Poškodba	Vzrok poškodbe	FES
80 kg	188 cm	27 let	nepopolna C6-7	skok v vodo na glavo	ne



Slika 6.1: Pacient LG. Primerjava izračunanih in izmerjenih vrednosti med senzornim sistemom in optičnim merilnim sistemom Optotrak®.



Slika 6.2: Pacient LG. Časovni poteki kinematičnih parametrov, kotne hitrosti golena, pospeška gležnja in kotov v gleženjskem in kolenskem sklepu. Dodan je tudi izhod algoritma za detekcijo faz in za ocenjevanje kvalitete zamaha, ki že upošteva napačno detektirane faze zamaha in vrne negativno vrednost.

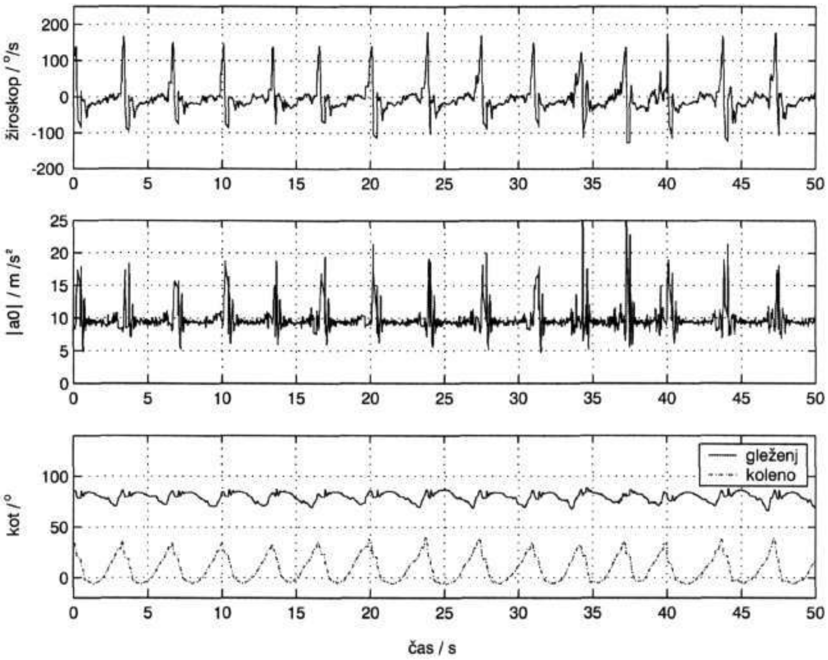
6.2 Rezultati meritev pri pacientu BM z nepopolno poškodbo C6

Pacient BM je bil prva testna oseba, ki je hodil po tekočem traku s prednastavljeno ustrezno hitrostjo 0.7 km/h. Meritve so bile izvedene v dveh fazah. Prva faza je potekala brez FES, čeprav je imel pacient težave s padanjem stopala. Referenco za kvaliteto zamaha smo zajeli med enako hojo na drugi nogi ob uporabi predpisanega lastnega stimulatorja MicroFES s petnim stikalom na levi nogi (na desni ni bilo stimulatorja). Tako je bila hoja, čeprav neučinkovita, bolje ocenjena. Na sliki 6.3 je lepo razvidna razlika med fazo opore in fazo zamaha. Pacient je precej silovito zadeval ob tla. Slika 6.4 prikazuje oceno faze zamaha. V primeru nepravilne detekcije zamaha, je sistem ocenil zamah z negativno vrednostjo. V drugi fazi smo uporabili računalniško voden stimulator (slika 4.1) z naslednjimi stimulacijskimi parametri: frekvenca stimulacije 20 Hz, amplituda 45 mA, pulzna širina 80 μs, proženje ročno s tipko. Pacient je bil preveč utrujen, kar je zaslediti v signalih pospeškometra (a_0) in žiroskopa kot upad amplitude. (slika 6.5 in 6.6).

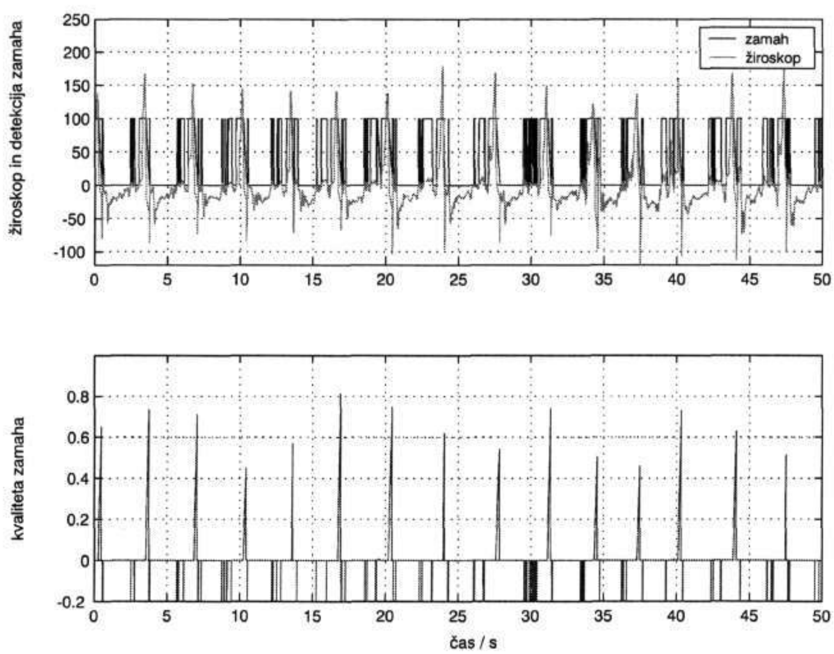
Tabela 6.2: Podatki pacienta BM.

Teža	Višina	Starost	Poškodba	Vzrok poškodbe	FES
68 kg	172 cm	38 let	nepopolna C6	prometna nesreča	MicroFES, 2 meseca

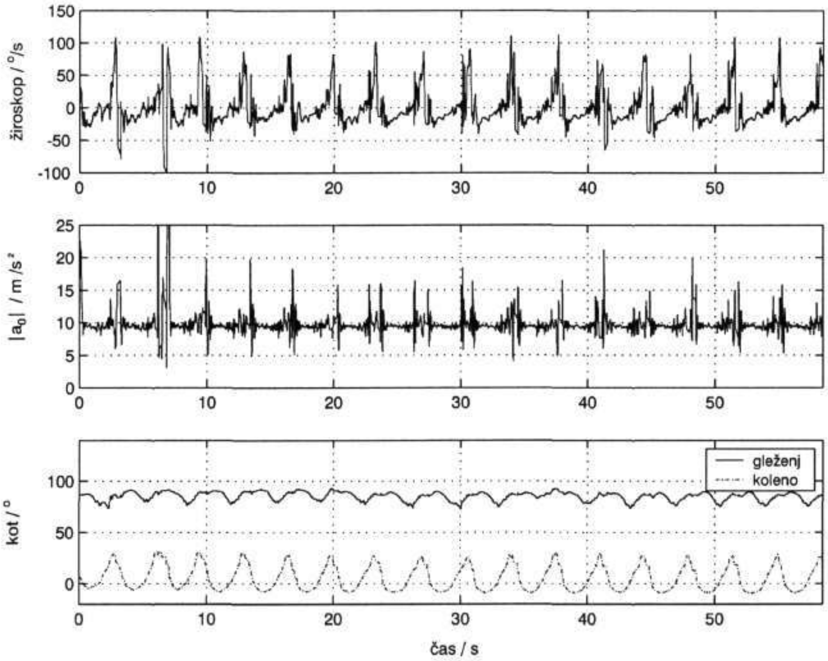
Meritve smo ponovili po enem tednu. V tem času je bil pacient deležen rehabilitacijskega treninga na tekočem traku z delno razbremenitvijo teže. Medtem je bila načrtana tudi kognitivna povratna zanka. V obliki zvočnih piskov treh različnih frekvenc je bila pacientu posredovana informacija o kvaliteti zamaha (dober, zadovoljiv, slab). Na podlagi posredovane informacije je hoteno vplival na svoj zamah, pri tem pa mu je pomagala FES (frekvenca stimulacije 20 Hz, amplituda 27-32 mA, pulzna širina 70 μs, proženje ročno s tipko), ki jo je prožila fizioterapevtka s tipko po lastni presoji in na podlagi kognitivne povratne zanke. Če je bila FES prožena v trenutku zamaha, je bil zamah dober. Tudi če je bila prožena malenkost prezgodaj, se je obnesla. Vendar smo v tem primeru dobili še dodatno zaznane faze zamaha, ki pa so bile napačne (slika 6.7). Tak pristop je sicer izboljšal rezultate in je hkrati privedel pacienta do zavestnega sodelovanja v rehabilitacijskem procesu, vendar analiza rezultatov kaže, da je za zagotavljanje ponovljivosti pri učenju hoje na tekočem traku potrebno zagotoviti ponovljivo proženje in trajanje električne stimulacije.



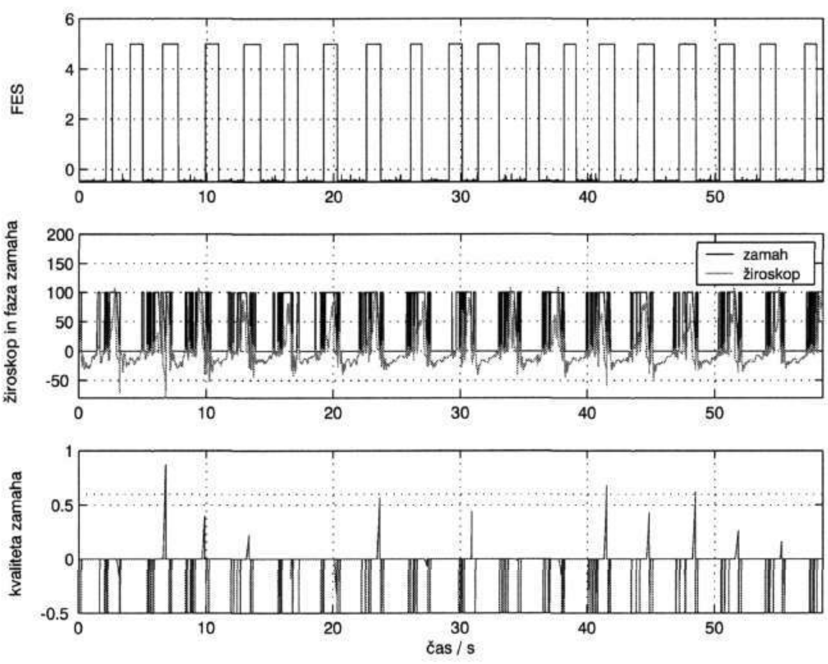
Slika 6.3: Pacient BM (22.11.2001 meritev 3). Časovni poteki kinematičnih parametrov, kotne hitrosti golena, pospeška gležnja in kotov v gleženjskem in kolenskem sklepu.



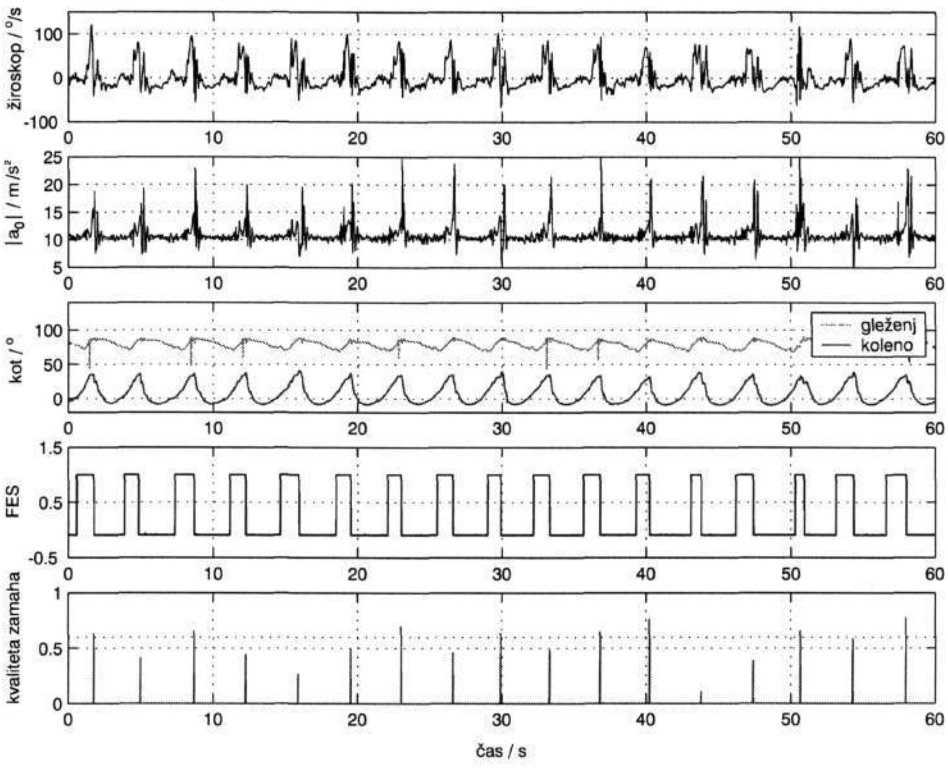
Slika 6.4: Pacient BM (22.11.2001 meritev 3). Kotna hitrost golena in detekcija faze zamaha. Kadar pride do napačne detekcije, to zazna algoritem za oceno kvalitete zamaha.



Slika 6.5: Pacient BM (22.11.2001 meritev 14). Časovni poteki kinematičnih parametrov, kotne hitrosti golena, pospeška gležnja in kotov v gleženjskem in kolenskem sklepu.



Slika 6.6: Pacient BM (22.11.2001 meritev 14). Kotna hitrost golena in detekcija faze zamaha. Kadar pride do napačne detekcije, to zazna algoritem za oceno kvalitete zamaha. Pacient je bil utrujen in ni dosegal želene dinamike pri hoji.



Slika 6.7: Pacient BM (29.11.2001). Časovni poteki kinematičnih parametrov. Graf prikazuje tudi prisotnost FESa in oceno kvalitete zamaha. Opazimo, da se trajanje FES spreminja od koraka do koraka, kar vpliva na kvaliteto zamahov in zmanjšuje ponovljivost procesa.

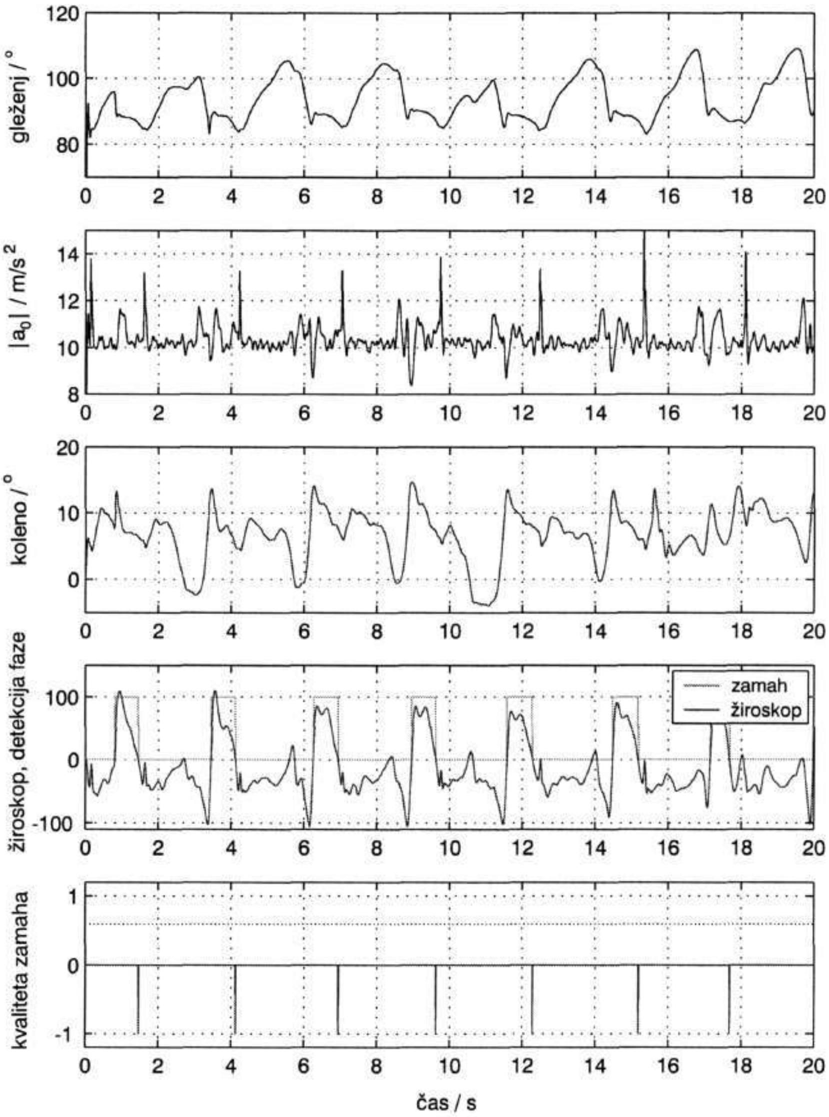
6.3 Rezultati meritev pri pacientu ID z zoženjem spinalnega kanala C2-C6

Pri pacientu ID smo naleteli na težavo, ki lahko pogosto spremlja osebe s poškodbo hrbtenjače. To je pomanjkanje fleksije v kolenskem sklepu zaradi možne atrofije fleksorjev kolena. Nezadostna kolenska fleksija povzroči nezmožnost izvajanja faze zamaha, zato je pacient prisiljen dvigniti kolk, da lahko noga kljub ekstenziji kolena napreduje in preide v fazo zamaha. Rešitev smo iskali v vključitvi novega, pomožnega, stimulacijskega kanala, ki v fazi opore stimulira fleksorje kolena (frekvenca stimulacije 20 Hz, amplituda 40 mA, pulzna širina 200 μs) in omogoči fleksijo kolena. Potem lahko sprožimo peronealno stimulacijo in posledično izzovemo fleksijski odziv, ki pomaga pacientu izvesti zamah.

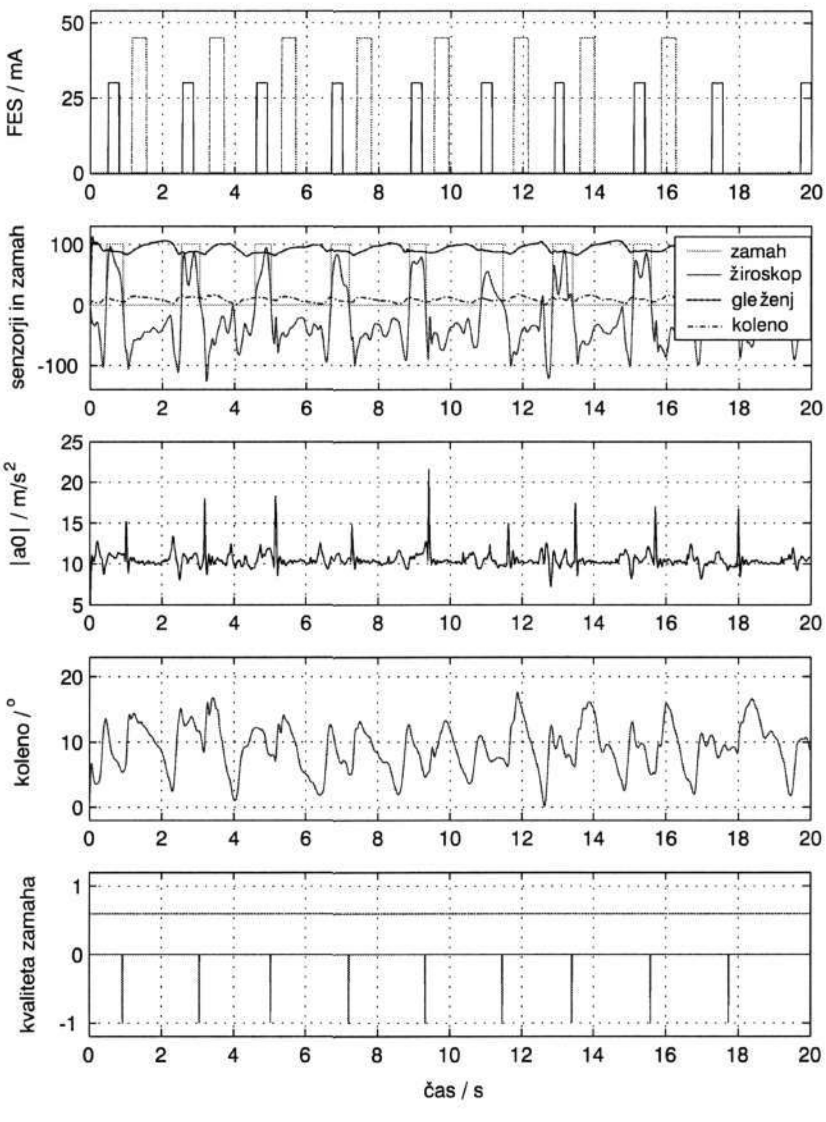
Tabela 6.3: Podatki pacienta ID.

Teža	Višina	Starost	Poškodba	Vzrok poškodbe	FES
88 kg	180 cm	65 let	C2-C6	Stenosis canalis spinalis	ne
			C5-C6	Protrusio disci	

Slika 6.8 prikazuje hojo pacienta brez FESa. V signalu pospeškometra je zaznati trd dotik s peto ob koncu faze zamaha, medtem ko v goniogramih zasledimo nezadostno fleksijo v kolenskem sklepu. Slika 6.9 prikazuje meritev z FES. Stimulacijo fleksorjev kolena proži konica, ki se pojavi v časovnem poteku pospeškometra ob dotiku pete s tlemi. Želen trenutek proženja nastavimo z dodatno zakasnitvijo. Fleksija kolena v fazi opore sprosti kolenske ektenzorje, kar omogoča fleksijo kolena pri fleksijskem odzivu, ki ga dobimo s peronealno stimulacijo. Peronealno FES (frekvenca stimulacije 20 Hz, amplituda 20 mA, pulzna širina 200 μs) proži kolenski goniometer, ko doseže predpisan kot v predfazi zamaha. Trenutek proženja dodatno nastavimo z zakasnitvijo. Izkazalo se je, da stimulacija peronealnega živca ni izzvala popolnega fleksijskega odziva, kar opazimo na goniogramu kolenskega sklepa. Pacient je bil že naučen hoje z dvigovanjem kolka, zato je tako nadaljeval, fleksija v kolenu je bila še vedno nezadostna, posledično pa trajanje peronealne FES prekratko in zamah slab.



Slika 6.8: Pacient ID (28.03.2002). Časovni poteki kinematičnih parametrov, kotne hitrosti golena, pospeška gležnja in kotov v gleženjskem in kolenskem sklepu. Opazimo zmanjšano dinamiko pospeška v gležnju in premajhno fleksijo kolena. Brez FESa.



Slika 6.9: Pacient ID (28.03.2002). Časovna razporeditev vlakov stimulacijskih pulzov. Stimulacija fleksorjev kolena (črtkano) nastopi v fazi opore, proži jo konica, ki se pojavi v časovnem poteku pospeškometra ob dotiku pete s tlemi. Sledi vlak pulzov peronealne stimulacije, ki jo sproži predpisana kolenska fleksija. Peronealna stimulacija ne izzove popolnega fleksijskega odziva. Pomanjkanje fleksije v kolenskem sklepu v fazi zamaha zahteva dodatno stimulacijo kolenskih fleksorjev. Posledično je bil tudi zamah slab.

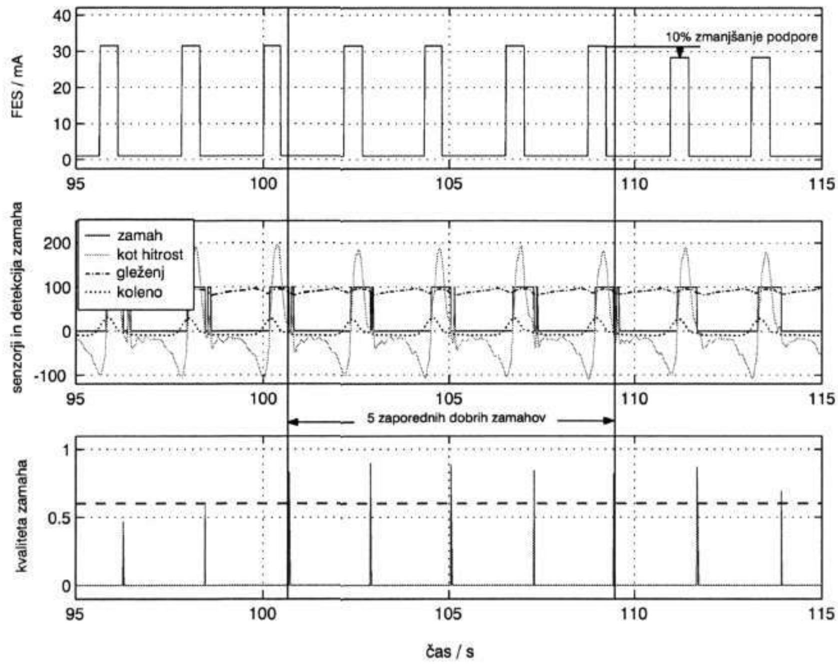
6.4 Rezultati meritev pri pacientu BK z nepopolno poškodbo C4-5

Meritve pri pacientu BK so potekale z novejšo različico algoritma za ocenjevanje zamaha, ki izloči nepravilno detektirane zamahe in tega ne posreduje v kognitivni povratni zanki. Novo je bilo tudi avtomatsko proženje FES in sistem za nadzor jakosti stimulacije. S tem smo uvedli možnost večjega hotenega sodelovanja pacienta v rehabilitacijskem procesu. Pacient je uporabnik MicroFESa, ki ga je uporabljal tudi med zajemanjem referenčnega zamaha pri hoji po tekočem traku. Nato je bil stimulator na desni nogi zamenjan z računalniško vodenim stimulatorjem (slika 4.1). Parametri stimulacije: frekvenca stimulacije 20 Hz, amplituda 35 mA, pulzna širina 200 μs, proženje avtomatsko na podlagi senzorne informacije.

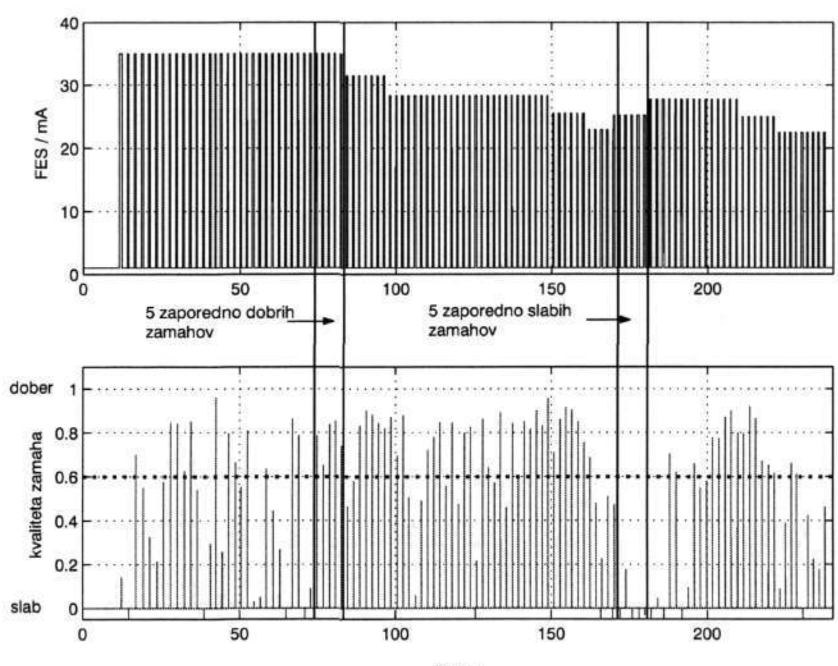
Tabela 6.4: Podatki pacienta BK.

Teža	Višina	Starost	Poškodba	Vzrok poškodbe	FES
84 kg	175 cm	30 let	nepopolna C4-C5	padec na ledu	MicroFES, 1 mesec

Slika 6.10 prikazuje uspešno izvajanje petih zaporednih dobrih zamahov. Ko sistem prešteje pet dobrih zamahov, zniža jakost električne stimulacije za 10% in s tem zmanjša motorično podporo pacientu in obratno, če prešteje pet slabih zamahov. Ob izmeničnem izvajanju dobrih ali slabih zamahov se FES motorična podpora ne spremeni. Število potrebnih zaporednih zamahov za spremembo FES motorične podpore je nastavljivo. Tako pacient nadaljuje hojo z zmanjšano/povečano FES motorično podporo. V primeru uspešnega vključevanja pacienta v proces, se jakost stimulacije zmanjšuje (slika 6.11).



Slika 6.10: Pacient BK (18.04.2002). Pet zaporedno dobro izvedenih zamahov povzroči znižanje motorične podpore, tj peronealne stimulacije, za 10% in obratno.



Slika 6.11: Pacient BK (18.04.2002). Kadar je pacient uspešen pri izvajanju zamahov, se motorična podpora niža in pacientovo sodelovanje ima vedno večji pomen. V primeru petih zaporednih slabih zamahov se nivo stimulacije ponovno zviša.

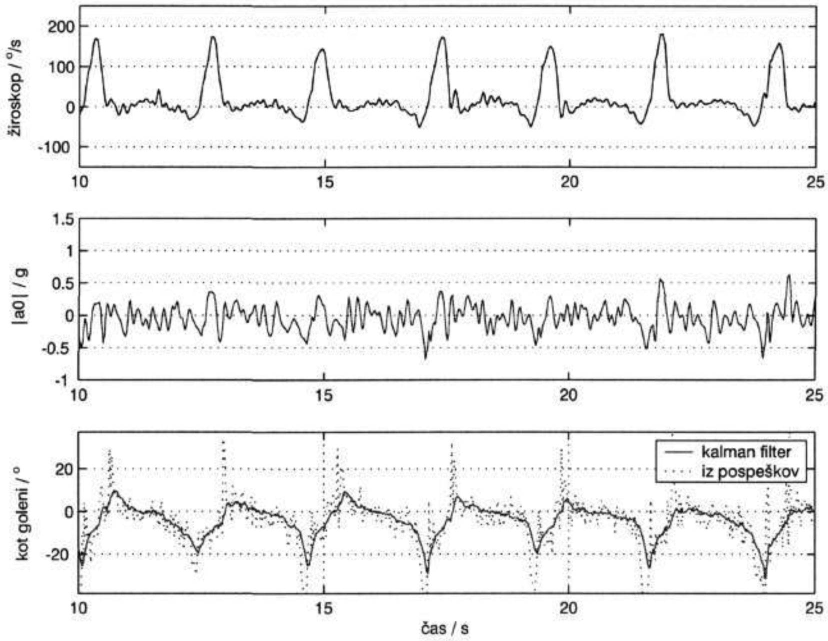
6.5 Rezultati meritev pri osebi ZM brez gibalnih motenj

Pri pacientu BK so bili doseženi zelo dobri rezultati, edino pomanjkljivost je bilo videti pri uporabi goniometrov. Njihova namestitev ni bila zanesljiva, še manj pa ponovljiva, saj se je zaradi pritrditve z velcro trakom spreminjal tudi položaj goniometra, kar je hkrati povzročalo neprijetnosti pacientu pri hoji. Goniometer smo nadomestili z razvitim algoritmom, ki je omogočal zanesljivo določevanje kota med golenom in vertikalno ravnino. Izračunana informacija je v nadaljevanju služila za vodenje električne stimulacije.

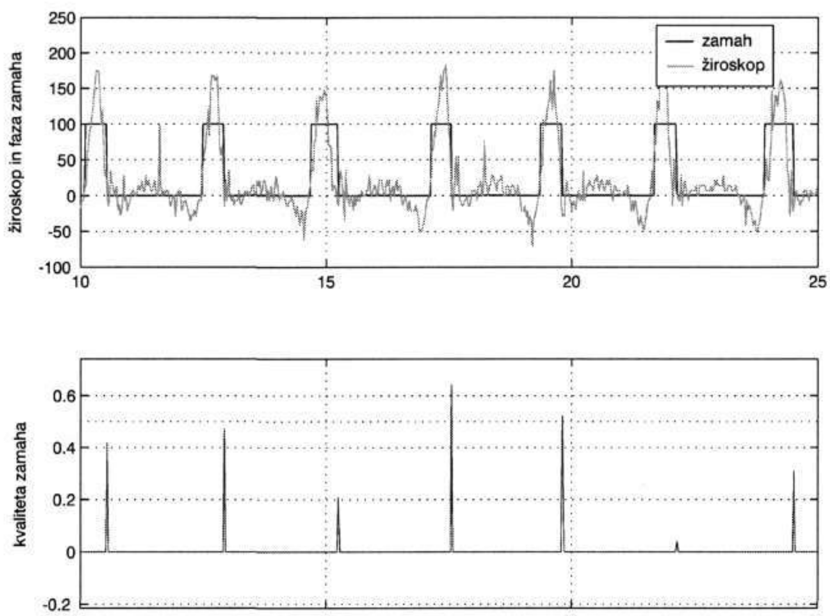
Tabela 6.5: Podatki osebe ZM.

Teža	Višina	Starost	Poškodba	Vzrok poškodbe	FES
95 kg	190 cm	35 let	brez gibalnih motenj		

Algoritem je bil preskušen na testni osebi ZM brez gibalnih motenj. Slika 6.12 prikazuje poleg signala žiroskopa, kotne hitrosti golena, in absolutnega pospeška v gležnju še kot golena, določen z integracijo informacij s pomočjo Kalmanovega filtra.



Slika 6.12: Oseba ZM (03.12.2002). Graf prikazuje kotno hitrost golena, informacijo, s katero detektiramo fazo zamaha ali jo uporabimo skupaj s tangencialnim in radialnim pospeškom pri izračunu Kalmanovega filtra za določanje kota golena.



Slika 6.13: Oseba ZM (03.12.2002). Detekcija faze zamaha in kvaliteta zamaha. Dober zamah (nad koeficientom 0.6) je enak referenčnemu, izmerjenemu v posebni meritvi.

6.6 Komentar rezultatov

Če naredimo povzetek vseh prikazanih rezultatov, naj poudarimo, da je bil njih cilj testirati tako posamezne algoritme, kakor tudi delovanje sistema kot celote. Hkrati je bil primarni cilj ugotavljanje pomanjkljivosti, ki so se pokazale med posameznimi testi. Pri pacientu LG smo preverili ali je smiselno sistem graditi na hoji po ravnem terenu ali se je bolje poslužiti že uveljavljenega tekočega traku. Hkrati je šlo za preverjanje delovanja senzornega sistema, ki naj bi nadomestil predrago in klinično nesprejemljivo optično opremo. Pacient BM je prvi preskusil delovanje algoritma za ocenjevanje kvalitete zamaha in kognitivne povratne zanke, ki je posredovala to informacijo v treh stopnjah. Pri nastavljeni hitrosti tekočega traku pacient ne bi bil sposoben razločiti med več stopnjami povratne zanke. Druga zanimivost, ki smo ji začeli posvečati pozornost, je proženje električne stimulacije. Čeprav se je trening hoje z ročnim proženjem izkazal za učinkovitega [86] pri hoji plegičnih oseb, pri paretičnih osebah ni bilo moč zagotoviti kontinuitete dobrih zamahov, saj stimulacija ni bila enaka. Različni so bili trenutki proženja glede na stanje okončin in tudi enakega trajanja ni bilo moč zagotoviti. Sledlje bi lahko zagotovili s prednastavljanjem časa trajanja, medtem ko je za zanesljivo in ponovljivo proženje bilo potrebno načrtati avtomatsko proženje s pomočjo senzorne informacije. To je bilo realizirano v primeru pacienta BK, ki je bil tudi ustrezni kandidat za ponovno učenje hoje.

Rezultati kažejo, da je možno z nadzorovano ponovljivostjo doseči izjemen napredek. Pri pacientu ID je šlo zgolj za možnost, da v sistemu predvidimo tudi uporabo drugega dodatnega stimulacijskega kanala. Nadaljni razvoj uporabe senzorne informacije pri ponovnem učenju hoje pa je prikazan pri osebi ZM, kjer je učinkovit algoritem izpodrinil uporabo nezanesljivih goniometrov.

7

Zaključek

V povzetku smo zapisali, da je hoja ena izmed najpomembnejših dejavnosti v vsakdanjem življenju. Velikokrat poškodba živčnega sistema ali možganska kap onemogoči normalno hojo ali jo ovira. V prvem primeru govorimo o hromi osebi, v drugem pa o osebi z gibalnimi motnjami.

Disertacija obsega raziskavo, ki je usmerjena v analizo hoje, tako normalne kakor motene. Z analizo hoje pridobimo informacije, na podlagi katerih ugotavljamo gibalne pomanjkljivosti. Predmet raziskave ni bila sinteza 'normalne' hoje, saj za popolno obnovitev plegične hoje nimamo na voljo ne zadostnega poznavanja fizioloških mehanizmov in niti ne tehničnih možnosti. Napredek, predstavljen iz vidika sinteze funkcionalne hoje paraplegičnih oseb, je bil relativno majhen, čeprav so bili prvi predstavljeni rezultati senzacionalni. Kralj je s sodelavci [3] predstavil štiritočkovno hojo paraplegikov, kar je sprožilo vsesplošno zanimanje za električno stimulacijo v paraplegiji [28]. Hkrati je bila električna stimulacija uporabljena v terapevtske namene, funkcionalni učinek pa je bil dosežen pri paretičnih osebah [82, 83]. Pri poškodbi hrbtenjače so ohranjeni naravni senzorji, prizadete pa so živčne poti in povezave, pri popolni poškodbi tudi prekinjene. Z ustrezno integracijo ohranjenih naravnih senzorjev in sistemov vodenja z umetno načrtanimi algoritmi aktivacije paraliziranih mišic je mogoče doseči nadzorovano hojo. Oseba ohrani nadzor nad hojo s hoteno aktivnostjo, umetni sistem vodenja pa v podrejenem položaju predstavlja uspešen rehabilitacijski pripomoček.

Raziskava se ukvarja z analizo in sintezo hoje pri paretičnih osebah. Temelj dosedanjih raziskav na področju hoje paretičnih oseb je bila neodvisna obravnava zavestne aktivnosti osebe ter umetnega sistema vodenja. Zavestna aktivnost osebe pri rehabilitaciji je bistvenega pomena za reorganizacijo osrednjega živčnega sistema [49]. Aktivni generatorji sile pri realizaciji gibanja in posledično hoje so skeletne mišice, ki jih nadzoruje živčni sistem. Slednji je adaptivna struktura, ki se prilagaja in razvija določene vzorce aktivnosti v obdobju filogeneze [88]. Stereotipični motorični vzorec, kot je hoja, se kaže v vzorcih mišične aktivnosti, le ti pa se generirajo avtomatsko v nasprotju z novo pridobljenimi vzorci, npr. športna aktivnost, ki jih osrednji živčni sistem dojema s ponavljanjem določene aktivnosti.

V primeru oseb z gibalnimi motnjami, bodisi zaradi bolezni ali poškodbe je naravna avtomatična gibalna aktivnost motena. Bistvo rehabilitacije je obnovitev motenih aktivnosti. V primeru popolne nefunkcionalnosti posameznih nevromišičnih skupin ali anatomskih sprememb je potrebno modificirati funkcijo ostalih, zdravih, nevromišičnih in skeletnih sistemov tako, da prevzamejo vlogo le teh. Skratka z reorganizacijo centralnega živčnega sistema dobimo nove vzorce nevromišičnih aktivnosti in posledično tudi nove vzorce hoje. Takega pristopa se poslužujemo tudi pri rehabilitaciji oseb z nepopolno poškodbo hrbtenjače. Z dolgotrajno vadbo razvije gibalno prizadeta oseba nov vzorec hoje. V ta namen moramo uporabiti ponovljive rehabilitacijske metode kot je hoja po tekočem traku z razbremenitvijo teže [52], uporaba lokomotornih naprav [58] ali robotskih mehanizmov za hojo [60].

V raziskavi smo se lotili novega pristopa, ki temelji na uporabi tekočega traku. Z uporabo umetnih senzorjev smo ovrednotili hojo in informacijo podajali osebi med hojo. Senzorna informacija tudi nadomesti subjektivno ocenjevanje hoje s strani fizioterapevtov. Hkrati pa opozarja pacienta na kvaliteto hoje in s tem izzove hoteno aktivnost.

V začetku smo se lotili predvsem analize hoje. Hojo smo razdelili v dve glavni fazi, fazo opore in fazo zamaha. S kinematičnim pristopom smo modelirali spodnjo okončino v obeh fazah. Z dinamičnim modeliranjem v fazi opore bi lahko določili potrebne momente v sklepih, ki bi jih generirali s funkcionalno električno stimulacijo [1], v kolikor bi se izkazala potreba po stimulaciji v fazi opore. V doktorski disertaciji smo se zaradi razsežnosti problema osredotočili na fazo zamaha. Na podlagi kinematične analize hoje smo določili posamezne faze hoje. V ta namen je bil razvit algoritem za detekcijo faze zamaha. Uporaba nevronskih mrež je bila ena izmed možnosti, ki se je ponujala kot rešitev tega problema. Vhodno - izhodno karakteristiko smo zapisali z večnivojsko mrežo nevronov, ki smo jih utežili tako, da so ustrezali seriji izvedenih meritev. Razvit je bil poseben algoritem, ki je na podlagi primerjave želenega in izmerjenega časovnega poteka pospeška v gležnju v času zamaha ob upoštevanju ustrezne fleksije v kolenu ovrednotil fazo zamaha. Za potrebe kognitivne povratne zanke je bilo potrebno informacijo še klasificirati tako, da jo je bil pacient sposoben razločiti pri sprejemljivi hitrosti tekočega traku.

Pri paretičnih osebah imamo opravka z motnjami motoričnih funkcij, zato nam je v veliko pomoč funkcionalna električna stimulacija, s katero obnovimo določene aktivnosti nevromišičnih skupin. Mirbagheri s sodelavci je pokazal, da dolgotrajna funkcionalna električna stimulacija zmanjša togost v sklepih in izboljša lokomotorne zmogljivosti osebe [89]. Ker je v rehabilitacijskem procesu zaželena hotena aktivnost paretične osebe, mora biti motorična podpora minimalna, torej le do nivoja, ki je potreben za izvedbo zamaha. Za obnovitev zamaha smo uporabili enokanalno peronealno stimulacijo. V prvi fazi smo se poslužili ročnega proženja, a se je izkazalo, da s tem ogrožamo ponovljivost rehabilitacijskega procesa. V ta namen smo avtomatizirali vodenje funkcionalne električne stimulacije. Proženje je bilo izvedeno na podlagi fleksije v kolenu, ki smo jo merili z elektrogoniometrom. Zahtevano minimalno motorično podporo smo zagotovili z nadzorom kvalitete zamaha. V primeru da je pacient opravil določeno število dobrih zamahov, je bil nivo motorične podpore znižan in obratno v primeru določenega števila slabih zamahov povišan. Če je izvedel manj kot pet slabih oz dobrih ali zadovoljivih zamahov, to ni imelo nobenega vpliva na nivo motorične podpore.

Kasneje se je izkazalo, da goniometri ovirajo pacienta, hkrati pa ostajajo najbolj kočljiv člen v sistemu in vnašajo vanj negotovost. Senzorno informacijo smo zato dopolnili z novimi informacijami, za katere so skrbeli algoritmi za določanje kota golena oz relativnega položaja spodnje okončine. Ker goniometri niso bili več prisotni v sistemu, je bilo proženje električne stimulacije osnovano na podlagi kota golena. Ko je slednji dosegel zahtevano vrednost v predfazi zamaha, je bil pacient deležen motorične podpore s peronealno električno stimulacijo.

Sistem za ponovno učenje hoje doprinaša veliko novosti v rehabilitacijo paretičnih oseb. Senzorna informacija, ki nastane z združevanjem, integracijo signalov iz različnih senzornih skupin, nam nudi veliko možnosti za analizo hoje, hkrati pa jo lahko uporabimo v raznih aplikacijah. Ena izmed takih je prav sistem za ponovno učenje hoje pri paretičnih osebah. Opisan pristop, ki ga predstavljamo, uspešno zagotavlja zahtevano ponovljivost v procesu rehabilitacije in hkrati zahteva hoteno sodelovanje pacienta. Na vidiku so dodatne izboljšave, ocenjevanje zamaha tudi v lateralni smeri, uporaba dodatnih senzorjev in združitev s sistemom, ki ocenjuje fazo opore. Glede na obetajoče rezultate, ki smo jih dobili s preliminarnimi meritvami, pa si bomo prizadevali izpeljati tudi obsežno klinično evalvacijo razvitega sistema, ki bi potrdila klinično uporabnost rehabilitacijskega pristopa.

Uporabljeni simboli

	Kinematika hoje
m_1, I_1, L_1	masa, vztrajnostni moment, dolžina stegna
m_2, I_2, L_2	masa, vztrajnostni moment, dolžina golena
m_3, I_3, L_3	masa, vztrajnostni moment, dolžina stopala
α	kot med horizontalo in stopalom
β	zasuk stopala pri hoji
γ	kot med stegnom in trupom
φ_1	kot kolenskega sklepa
φ_0	kot gleženjskega sklepa
x_p, y_p, z_p	prijemališce sile na podlago
F	sila na podlago
$T_1 ... T_5$	koordinate markerjev optičnega merilnega sistema
a_{t0opt}, a_{r0opt}	tangencialni in radialni pospešek v gležnju, izračunan iz optičnega merilnega sistema
$\|a_{0opt}\|, \|a_0\|$	absolutna vrednost pospeška v gležnju; iz optičnega in večsenzornega merilnega sistema
T_s	vzorčni čas
	Določanje faze zamaha, uporaba nevronskih mrež
$u(t)$	vhodni signal za identifikacijo nevronske mreže
$y(t)$	izhodni signal za identifikacijo nevronske mreže
$\varphi(t)$	regresijski vektor
$g(\varphi)$	nelinearna preslikava posameznega nevrona
n_a	število preteklih izhodov
n_b	število preteklih vhodov
n_k	zakasnitev sistema

$\hat{y}(t)$	izhod nevronske mreže
Z^N	niz podatkov za učenje mreže
Φ	uteži nevronske mreže
ω	kotna hitrost golena
	Kvantitativna ocena faze zamaha
$r1$	razdalja med gleženjskim sklepom in prvim parom pospeškometrov
$r2$	razdalja med gleženjskim sklepom in drugim parom pospeškometrov
a_{t0}	tangencialni pospešek
a_{r0}	radialni pospešek
θ_1	kot kolenskega sklepa
ω	kotna hitrost golena
T	vzorčni čas
	Kalmanov filter in ocena kota golena
$\mathbf{K}_{Kal}$	ojačenje Kalmanovega filtra
$\mathbf{P}$	kovariančna matrika Kalmanovega filtra
$\mathbf{x}_k$	predstavlja vektor vseh stanj
$\hat{\mathbf{x}}_k$	predstavlja vektor vseh ocenjenih stanj
$\mathbf{z}_k$	predstavlja vektor vseh merjenih signalov
ϕ	matriko prehajanja stanj
$\mathbf{B}$	vhodno matriko
$\mathbf{H}$	izhodno matriko
$\mathsf{Q}_{\mathbf{w}k}$	šum sistema
$\mathsf{R}_{\mathbf{v}k}$	merilni šum
T_s	vzorčni čas
$\Delta\hat{\theta}$	ocenjena napaka kota golena glede na vertikalo
θ_1	kot kolenskega sklepa
g	gravitacijski pospešek
θ	kot golena glede na vertikalo
θ_m	izračunan kot golena glede na vertikalo
$\hat{\theta}$	ocenjen kot golena glede na vertikalo

$\Theta_m(s)$	kot golena glede na vertikalo v Laplaceovem prostoru
ω	kotna hitrost golena
$\Omega(s)$	kotna hitrost golena v Laplaceovem prostoru
$\mathbf{G}(s)$	prenosna funkcija VP filtra v Laplaceovem prostoru
	Določanje relativnega položaja spodnje ekstremitete
x_t	x koordinata v k.s. gleženjskega sklepa
z_r	z koordinata v k.s. gleženjskega sklepa
x_b	x koordinata v baznem k.s.
z_b	z koordinata v baznem k.s.
θ	kot golena glede na vertikalo
g	gravitacijski pospešek
ω	kotna hitrost golena
	FES
T_{pd}	zakasnitveni čas peronealne stimulacije
T_{ps}	čas trajanja peronealne stimulacije

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Dodatek A

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Use of telekinesthetic feedback in walking assisted by functional electrical stimulation

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A telekinesthetic feedback implemented into functional electrical stimulation (FES) orthosis is described. Single channel FES is used to provoke ankle dorsiflexion during walking. FES is controlled manually by a special lever, built into the handle of the crutch. The angular position of the lever defines the intensity of stimulation and thus the magnitude of the ankle dorsiflexion. The measured joint angle provides the feedback information about the ankle joint position, which is presented to the user as a force feedback applied to the control lever. As the first step in the development of a complex micromechatronic device, a simulated testing environment was prepared. A computer model, comprising dynamic foot characteristics, as well as agonistic and antagonistic muscle groups, substitutes the ankle joint. The model also includes fatiguing of the electrically stimulated muscles. For experimental purposes an actuated control lever was built. The efficacy of the telekinesthetic feedback was evaluated in a group of six healthy persons.

Introduction

Walking is one of the most important human activities. Most spinal cord injured (SCI) persons, however, are deprived of this advantage. In some cases their locomotion is severely disturbed or they are not able to walk at all. With the assistance of crutches and functional electrical stimulation (FES) standing and simple gait were restored [1].

Because of the absence of sensory feedback from the paralysed extremity, human vision remains the only useful feedback for a SCI walker. This of course interferes with the primary function of the vision which is observing the surroundings. The simplest command sensor, independent of human vision, is the heel switch used for FES assisted walking in hemiplegic patients [2]. The switch is located in the sole of the shoe on the affected side. When a patient voluntarily raises the heel of the affected leg, the heel switch triggers the stimulator, causing foot dorsiflexion during the swing phase of the gait.

The great majority of the incomplete SCI subjects, who are candidates for a FES orthosis, are crutch users [3]. Because of frequent malfunctioning of the heel switch, a pushbutton built into the handle of a crutch was introduced [4]. In this application, as long as the subject was depressing the pushbutton, the stimulation was applied to the electrodes placed over the ankle dorsiflexors. In this way, the FES assisted swing phase of walking was produced in incomplete SCI persons.

A major issue associated with the FES is the reduction of muscle force in time, as a result of fatigue. In electrically induced contractions, the muscle fatigue is essentially peripheral and presents a severe problem, particularly in those cases where detection of fatigue is not possible due to the absence of sensory feedback [5]. A possible way to compensate for this difficulty is to replace the pushbutton by a proportional control built into the handle of a crutch in a form of a lever. In this way the subject can increase the amplitude of the electrical stimulation and thus overcome the effect of muscle fatigue.

The purpose of the paper is to present an improved FES orthotic system with manual control lever and telekinesthetic feedback built into the handle of the crutch. The aim of the telekinesthetic feedback is to provide information about the position of the paralysed joint to the patient. The ankle joint angle is measured by an electrogoniometer and transferred to a torque motor actuating the axis of the control lever. This way, the resistance of the control lever to manual voluntary control is proportional to the magnitude of the joint angle. By adjusting the pressure on the control lever, the walking subject can overcome the problems of fatiguing of electrically stimulated muscles and also can deal with the obstacles on the ground.

Methods

FES orthosis

The basic concept of the described FES orthosis including the telekinesthetic feedback consists of the following components (figure 1):

(1) control unit with electrical stimulator;

(2) stimulation electrodes;

(3) manual control lever with electrical motor;

(4) goniometer.

The manual control lever is built into the handle of the crutch and is connected to the shaft of an electrical motor directly or through a gearbox. The user determines the amplitude of electrical stimulation by changing the lever rotation angle. The measured angle

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represents the input information to the control unit where the corresponding FES amplitude is computed. The stimulator, integrated into the control unit, uses the computed data and stimulates the ankle joint dorsiflexors through surface stimulation electrodes. The resulting dorsiflexion is measured by a goniometer, placed at the ankle. The magnitude of the joint angle is translated into the counteracting torque in the manual control lever, produced by a torque motor installed in the handle in the crutch.

In a preliminary study the FES orthotic system was replaced by a simulated environment. For the purpose of variable amplitude FES triggering and telekinesthetic feedback an experimental hardware unit was built (figure 2). The goniometer, the electrical motor and the control lever are replaced by a much bigger motor with a steering handle and an optical encoder. The biomechanical computer model of the human ankle replaces the real human lower extremity and is software generated. The control unit with the electrical stimulator is software generated as well.

The functional electrical stimulator is presented by a pulse generator having a variable amplitude output and

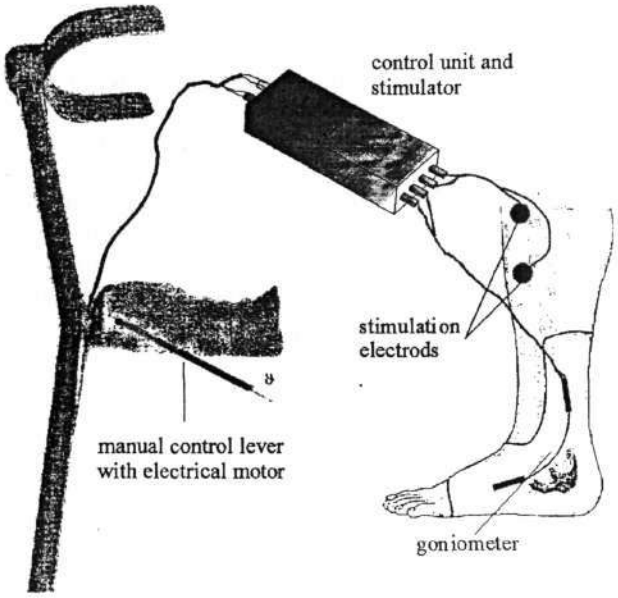


Figure 1. FES orthosis based on telekinesthetic feedback.

serves as the input into the musculoskeletal model of the human ankle. The amplitude of electrical stimulation is proportional to the control lever angle. Figure 3 shows two control loops. The first loop is used for returning the manual control lever into the starting position corresponding to the neutral position of the ankle joint. This loop has no special effect on the torque feedback; in this way the user develops a feel for the floating zero. The starting point is defined by the input parameter ϑ_{ref} and is set by the user. The second loop controls the counteracting torque. The output of the musculoskeletal model, the ankle angle, is turned into a control signal delivered to the electrical motor by the control unit.

Musculoskeletal model

The ankle joint model includes several independent subsystems: the ankle muscle groups, the joint characteristics and the foot being part of a whole musculoskeletal system. The ankle joint muscles are divided into two muscle groups, performing the opposite tasks: the agonistic muscle group performs the dorsal foot flexion and the antagonistic muscle group performs the plantar foot flexion.

The muscle group characteristic parameters are presented as viscous damping $B_1 (B_2)$, inner moment of inertia $J_1 (J_2)$ and elasticity $k_1 (k_2)$. The index 1 belongs to agonist while 2 to the antagonist muscle group. The foot moment of inertia J_3, the joint elasticity k_3 and the joint viscous damping B_3 supplement the model in conditions of free movement. The moment of inertia of the foot was calculated by replacing the foot with a prism [6]. The following equations describe the behaviour of the ankle joint in unconstrained conditions (1):

$$\begin{aligned} J_1\ddot{\varphi}_1 &= -B_1\dot{\varphi}_1 + T_{iso} - k_1(\varphi_1 - \phi), \\ J_3\ddot{\phi} &= -k_1(\phi - \varphi_1) - k_3\phi - B_3\dot{\phi} - k_2(\phi - \varphi_2), \\ J_2\ddot{\varphi}_2 &= -k_2(\varphi_2 - \phi) - B_2\dot{\varphi}_2, \end{aligned} \tag{1}$$

where ϕ represents the ankle joint angle, while φ_1 and φ_2 are the displacement of the agonist and antagonist muscle respectively.

The dorsiflexor moment generator is denoted by T_{iso}. It only acts in the direction of muscle contraction. The moment generator T_{iso} is not linearly proportional to

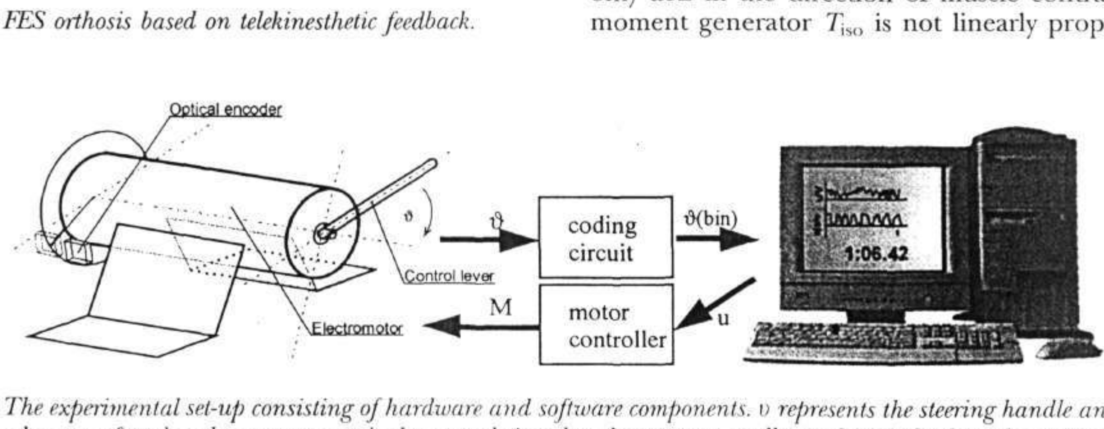


Figure 2. The experimental set-up consisting of hardware and software components. ϑ represents the steering handle angle, $\vartheta(bin)$ is the binary value transferred to the computer. u is the control signal to the motor controller and M is the desired counteracting torque, produced by the electrical motor.

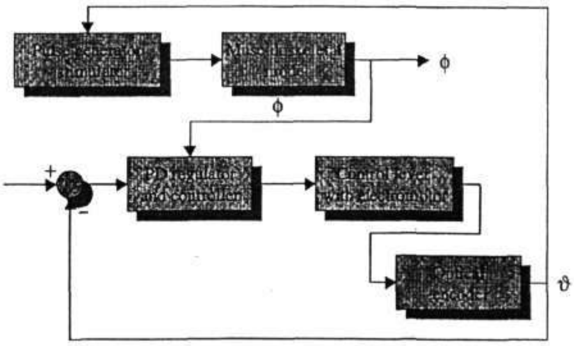


Figure 3. Block diagram of the system. υ is the angle of the manual control lever while ϕ represents the ankle joint angle. The starting point of the steering handle is defined by the input parameter υ_{ref} and is set by the user.

the stimulating voltage U_{st}. Two nonlinearities are incorporated, the threshold stimulation voltage and the saturation voltage [6]. T_{iso} is also a time-dependent function, defined by the muscular fatigue.

The numerical values of these elements cannot be determined by a direct measuring method. For that purpose frequency response based methods and identification approaches were used [6] (k_1=3 mN rad^{-1}, J_1=0.0245 mN s^2 rad^{-1}, B_1=0.17 mN s rad^{-1}, k_2=3 mN rad^{-1}, J_2=0.0245 mN s^2 rad^{-1}, B_2=0.17 mN s rad^{-1}, k_3=13 mN rad^{-1}, J_3=0.024 mN s^2 rad^{-1}, B_3= 2.7 mN s rad^{-1}).

Fatigue in muscles activated by FES

In general, the fatigue is the inability of a muscle or muscle group to continuously produce the required force, regardless of the type of work in progress. Muscular fatigue is a time-dependent process that may take place gradually or abruptly [5]. The fatigue can also be considered as a protective mechanism of the body to prevent permanent muscle injuries.

The nature of fatiguing observed in paralysed muscles activated by FES is different from normal fatigue. If the comparison with voluntary activated normal muscle fatigue is made, the following differences can be noticed [5]:

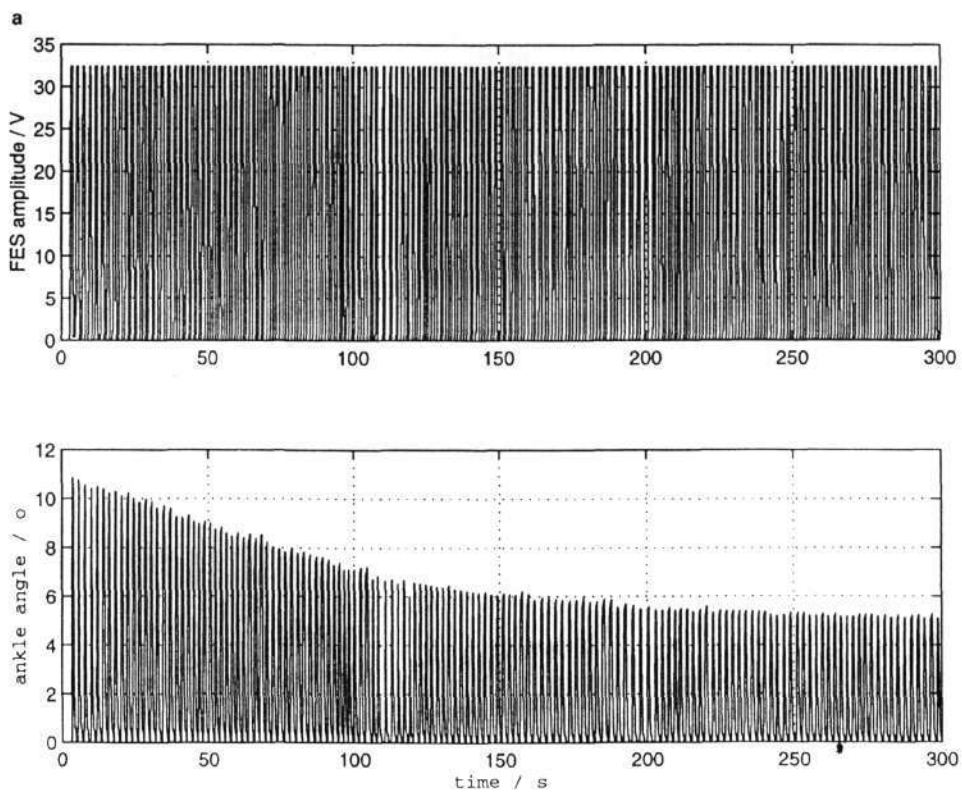
(1) mode of activation: FES synchronously activates all muscle motor units;

(2) reverse recruitment order: the rapidly fatiguing fast motor units are recruited at low stimulus intensities;

(3) peripheral fatigue: the central component of fatigue is missing because of the spinal cord injury;

(4) lack of sensory feedback: the muscle fatigue is not percepted by pain.

Electrically stimulated muscles fatigue more rapidly than when voluntarily activated. With intermittent stimulation,



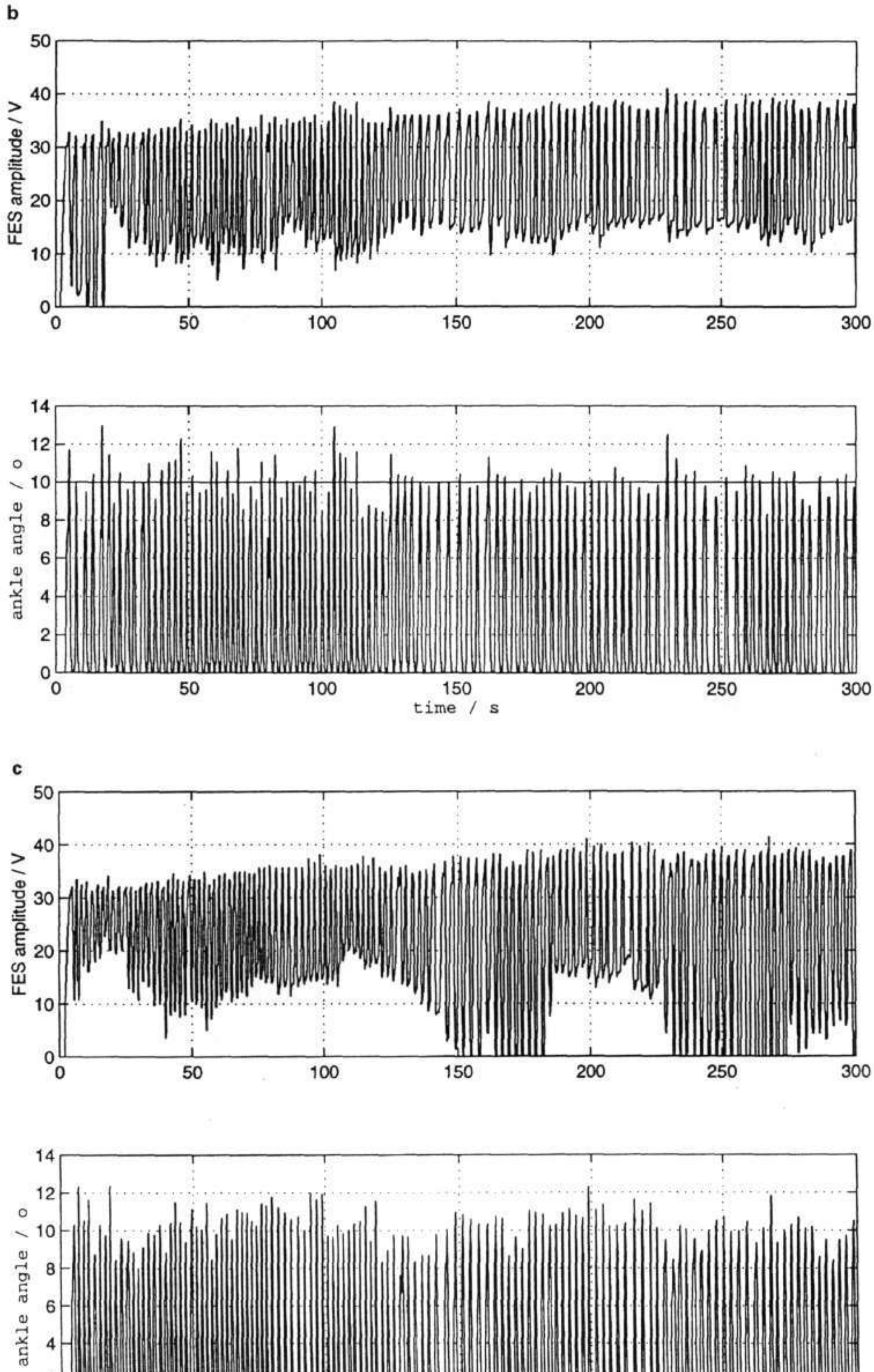


Figure 4. *(a) The upper diagram shows the trains of electrical stimulation while the lower diagram presents the effect of the muscular fatigue on the ankle goniogram. (b) In the second test the tested person was trying to reach the desired ankle joint angle (10°) while using both visual and telekinesthetic feedback. The effect of the muscular fatigue was overcome by increasing the stimulation amplitude (upper diagram). (c) The relationship between the ankle joint angle and the counteracting moment in the control lever was the only feedback information in the third test. The lower diagram presents a successful attempt of using the telekinesthetic feedback.*

Table 1

	Test									
Test	1		2		3		4		5	
Subject	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i	$\overline{x}_i$	s_i
1	9.85	1.19	9.94	1.22	10.1	1.67	9.15	1.75	9.99	1.71
2	10.6	1.29	10.5	1.27	11.1	1.41	10.6	1.47	10.3	1.53
3	10.3	1.59	10.9	1.16	9.52	1.66	10.1	1.38	10.9	0.99
4	10.3	1.83	10.7	1.74	8.88	1.68	11.1	1.02	10.3	1.39
5	8.70	0.89	9.08	1.46	8.92	1.05	9.67	1.23	10.7	0.97
6	10.3	1.19	8.62	1.08	9.50	1.25	9.80	1.26	10.6	1.43

where trains of stimuli are followed by a pause, fatiguing of electrically stimulated muscle is considerably less when compared to continuously delivered stimulation.

For the purpose of muscle fatigue modelling the following time course of the muscle moment was selected [5]:

$$T_{\text{iso}} = \{a_0(1 - a_1 \exp^{(-t/\tau_1)} - a_2 \exp^{(-t/\tau_2)}) - a_3 \tanh[(t - t_3)/\tau_3]\}T'_{\text{iso},} \quad (2)$$

where T'_{iso} is the moment generator with excluded muscular fatigue.

The parameters were identified from the measurements performed with intermittent stimulation where a train of pulses and the pauses lasted one second each [1]. On the basis of the comparison with curve (2) the following numerical values were obtained: $a_0 = 1$, $a_1 = -0.95$, $a_2 = 1$, $a_3 = 0.55$, $\tau_1 = 60$, $\tau_2 = 60$, $t_3 = 8$, $\tau_3 = 120$. Such a stimulation pattern to some extent corresponds to slow walking which is often exhibited by severely paralysed SCI subject.

Instrumentation

During experiments, a personal computer (PC) Pentium™ with data acquisition board AD512 from Humusoft s.r.o. was used. As a programming tool, Matlab® with the Simulink toolbox from The MathWorks Inc. was selected. The program uses block programming with built-in mathematical functions and makes possible the implementation of the control and communication tasks via C written S-functions. The complete musculoskeletal model including muscular fatigue was realized using block programming.

The control lever of the telekinesthetic feedback system was directly connected to the electromotor (dc electromotor from Minertia® Motor, Yaskawa Electric Japan) providing the counteracting torque. An optical incremental encoder was used for position information of the control lever. The encoder output was in standard quadrature signal output form; A,B impulses, phase shifted by $\pi/2$, and index signal R, defining the home position of the control lever. These output signals were converted into a parallel 12 bit signal by external hardware.

Results

Six unskilled users were selected to test the designed telekinesthetic feedback system. They were asked to repeatedly actuate the control lever in a way similar to walking. The simulated gait was divided in two phases, swing and stance. During the swing phase, the simulated muscle was electrically stimulated for the duration of approximately one second. The same duration was used for muscle relaxation during the stance phase of the virtual walking. The computer displayed the time in seconds (figure 2). The subject triggered the train of pulses by moving the control lever. Discontinuation of FES was achieved by releasing the handle. The duration of each test was five minutes.

In the first test electrical stimuli of constant amplitude were applied to the simulated ankle dorsiflexors. The influence of fatiguing of electrically stimulated muscle is noticeable (figure 4 (*a*)). This pattern corresponds to the stimulation responses provoked by the presently used drop-foot electrical stimulators.

In the second test the visual feedback was included. The visual feedback was presented as the time course of the computer simulated ankle angle displayed on the computer screen while the telekinesthetic feedback was simultaneously providing the counteracting torque. The displayed diagram comprised the desired peak value of the ankle joint angle (the line in figure 4 (*b*)). The subjects were asked to keep the maximal value of the ankle angle as constant as possible (10° of ankle dorsiflexion). The inherent muscular fatigue made this attempt difficult. Because of the muscular fatigue, the ankle angle was decreased and correspondingly also the counteracting moment of the control lever. This second test was also considered a learning phase for the third test. The users were taught about the relation between ankle joint angle and the counteracting torque.

The third test presents the behaviour of the telekinesthetic feedback system alone. A walking subject should be able to control the ankle joint dorsiflexion and use the telekinesthetic feedback information without the visual feedback. A skilled user (after two hours of training) was able to control the ankle joint angle within a range of $\pm 2°$ during the cyclical stimulation pattern which is more than satisfactory regarding the requirements for a FES orthosis (figure 4 (*c*)). Table 1 presents the results gathered in six healthy persons using the telekinesthetic feedback only. The main goal was to maintain the maximal value of the ankle angle (10°) during the five minutes test as constant as possible. For every person an average value and a standard deviation were calculated. The test was performed in five trials to achieve higher statistical reliability.

Conclusions

The telekinesthetic feedback was found efficient for the elimination of the effects of muscular fatigue during simulated gait of spinal cord injured persons. The muscular fatigue decreases the effect of FES without the user's awareness. When using the telekinesthetic feedback the user becomes aware of the fatigue and can increase the FES amplitude to reduce its influence.

In this paper the idea of the use of telekinesthetic feedback was presented on the example of the ankle joint FES control. The telekinesthetic feedback could be applied in several other applications. One of them is FES assisted hand grasping [7]. Here the hand control is achieved by a control lever attached to the shoulder. In this case the telekinesthetic feedback could be applied to the grasping force.

The use of the telekinesthetic feedback is helpful not only when dealing with the fatigue of the electrically stimulated muscles. It can also diminish the variability in responses of stimulated muscles caused by inaccurate daily positioning of surface electrodes. Furthermore, the telekinesthetic feedback can be found useful when walking over uneven terrain.

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Dodatek B

FES Rehabilitative Systems for Re-Education of Walking in Incomplete Spinal Cord Injured Persons

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Abstract

Objective. The aim of the paper is to present various relatively simple functional electrical stimulation (FES) systems that affect neural circuits and reflex behavior by providing necessary peripheral input to the lower extremities of incomplete spinal cord injured (SCI) persons.

Methods. The proposed FES re-education walking systems make use of feedback information that is transmitted from the paralyzed limb to the nonparalyzed part of the patient's body. A single gait variable can be analogously transmitted to the walking subject in a form of sensory stimulation. The information about several gait variables can be first integrated and afterwards delivered to the walking subject as a single command.

Conclusions. Significant improvements in the duration of the double support phase, metabolic energy expenditure, and physiologic cost index were observed when using FES-assisted training of walking in incomplete SCI persons. ■

Key Words: functional electrical stimulation, gait, incomplete spinal cord injury.

INTRODUCTION

It is a general observation that each year more incomplete spinal cord injured (SCI) patients are arriving in spinal units. Among them there are more incomplete tetraplegic than paraplegic cases. About one half of the incomplete SCI patients recover and need no orthotic aid after leaving the rehabilitation center. Functional electrical stimulation (FES) can be used as a therapeutic treatment in the early post-trauma phase within a great majority of the incomplete SCI cases(1,2).

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FES represents one of the rare rehabilitative approaches restoring walking patterns in stroke(3) and incompletely paralyzed SCI subjects soon after the accident(4). The aim of an FES rehabilitative system for re-education of walking, which we are proposing in this paper, is not only to deliver electrical stimulation to the paralyzed muscles, but also to assess the sensory information from the paralyzed limb. In order to foster the re-education process, the electrical stimulation should be voluntarily controlled by the patient where the sensory information is fed back to the patient and not to the stimulator control unit. These systems are intended to be used in incomplete SCI persons soon after the accident or onset of disease. The re-education FES systems are to be used within the rehabilitation centers and applied by therapists. Surface electrical stimulation is therefore appropriate. The sensory information can be assessed by the use of artificial sensors attached to the paralyzed limbs, e.g., goniometers, accelerometers, foot-switches, pressure insoles. This information can be delivered to the walking subject through electrical stimulation, tactile stimulation, audio signal, or haptic interface. In this paper we propose two of such FES re-education systems. In the first case the information about the joint angle is analogous to the force delivered to the patient through a haptic interface. In the second case the information from several artificial sensors is first integrated and afterwards delivered to the patient through sensory electrical stimulation.

ANALOG SENSORY FEEDBACK SYSTEM

In FES-assisted walking of SCI patients, vision represents the only useful feedback. However, vision should be primarily used for observing the surroundings and not to control the movement of the lower extremities. To overcome this disadvantage an analog sensory feedback system was developed. This system provides information to the user in the form analogous to the measured gait variable. This gait variable is usually assessed by the use of various sensors, mounted on the segments of the lower extremities in order to replace the natural sensory feedback. The problem arises how to present the feedback information to the walking SCI subject. One of the possible solutions is to use the haptic interface, providing a force feedback to the user. A haptic interface is known as a force-to-force feedback from the area of robotics where the end-effector force is transmitted to the steering handle(5). In general the force feedback can be analogous to any other kind of sensory information.

The FES re-educational systems for walking of incomplete SCI subjects with the assistance of crutches can make use of various sensors in order to determine position of the lower extremities, velocity, acceleration, or ground reaction forces. The proposed analog FES system uses a single goniometer to provide the information on ankle joint angle. The original FES orthotic aid(2) comprised only a pushbutton to trigger voluntarily the stimulation of the peroneal nerve. Mounting a goniometer to the subject's ankle joint causes a demand for a special steering lever that can provide a force feedback. The proposed FES re-educational system helps SCI subjects to "feel" the ankle joint angle in a form of a counteracting torque acting on the control lever(6) during each walking cycle.

A haptic interface for the FES orthosis (Fig. 1) consists of the following parts: control unit including the peroneal stimulator, stimulation electrodes, control lever with electrical motor, and goniometer. The user defines the intensity of electrical stimuli by changing the angular position of the control lever built in the handle of the crutch. The control unit computes the corresponding FES amplitude and stimulates the ankle dorsiflexors using the surface stimulation electrodes. The resulting ankle dorsiflexion is measured by the goniometer attached to the ankle joint. The measured angle is reflected as a counteracting torque in the hand control lever. The torque is produced by the electrical motor actuating the axis of the control handle.

The viability of the proposed analog sensory feedback system was evaluated by using computer modeling. In the preliminary investigation some of the miniature hardware parts were replaced by larger units. A large electrical motor with a control lever was built to serve as a haptic interface and to replace the miniature control lever, which was to be built together with the motor into the crutch handle. The biomechanical model of the human ankle replaced the human lower extremity and was software-generated. Its output was the ankle joint angle. Fatiguing of the electrically stimulated muscle was included

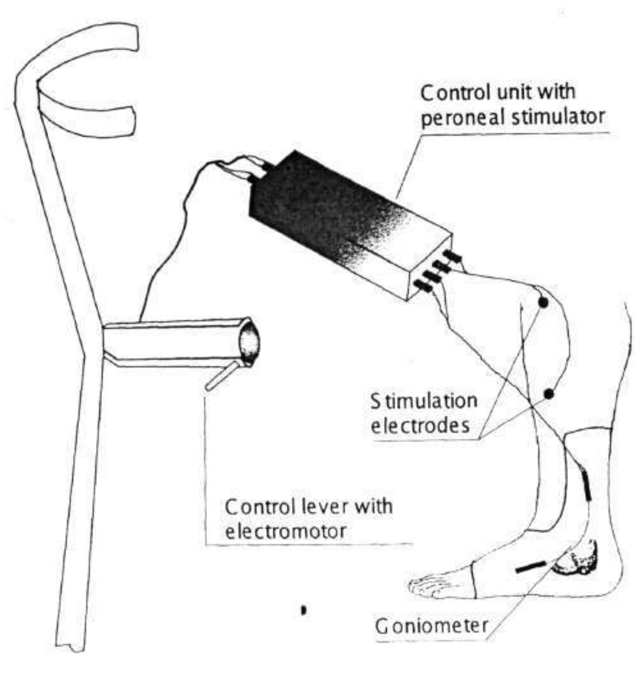

Figure 1. FES peroneal orthosis based on haptic interface.

into the muscle model. The electrical stimulator and the control unit were both simulated as software models. The first one was represented by a software pulse generator and was input to the biomechanical model(6) of the ankle joint, the second one was a controller calculating the control voltage proportional to the ankle angle. A voltage-controlled electrical motor provided the counteracting torque to the lever and consequently to the user.

Testing of the haptic interface experimental setup was carried out in six unskilled users. Their task was to actuate repeatedly the control lever in a similar way as during walking. The simulated gait was divided into two phases. During the swing phase the dorsiflexors were electrically stimulated for the duration of approximately one second, while during the stance phase of about the same duration the muscle relaxation occurred. The events of virtual walking were assessed by the computer and displayed on the screen. The duration of each test was five minutes.

In the first test (Fig. 2a) a train of electrical stimuli with constant amplitude was applied to the simulated ankle dorsiflexors. At this point it was our aim to demonstrate the problem of muscle fatigue occurring with the presently used stimulation systems. The second test included visual feedback in the form of the time course of the simulated ankle angle displayed on the computer screen, while the haptic interface was simultaneously providing the counteracting torque. The displayed diagram comprised also the desired peak value of the ankle joint angle (the line in Fig. 2b). The subject's goal was to maintain the maximal value of the ankle angle as constant as possible at the angle of 10° of ankle dorsiflexion. The inherent muscular fatigue was the obstructive factor, decreasing the ankle angle. This test was considered as training for the use of the haptic interface without visual feedback. The third test was a real evaluation of the haptic interface. The subjects were asked to control the ankle joint dorsiflexion by the use of the haptic interface information without any other feedback. A two-hour training made possible to control the ankle angle within a range of ± 2° (Fig. 2c).

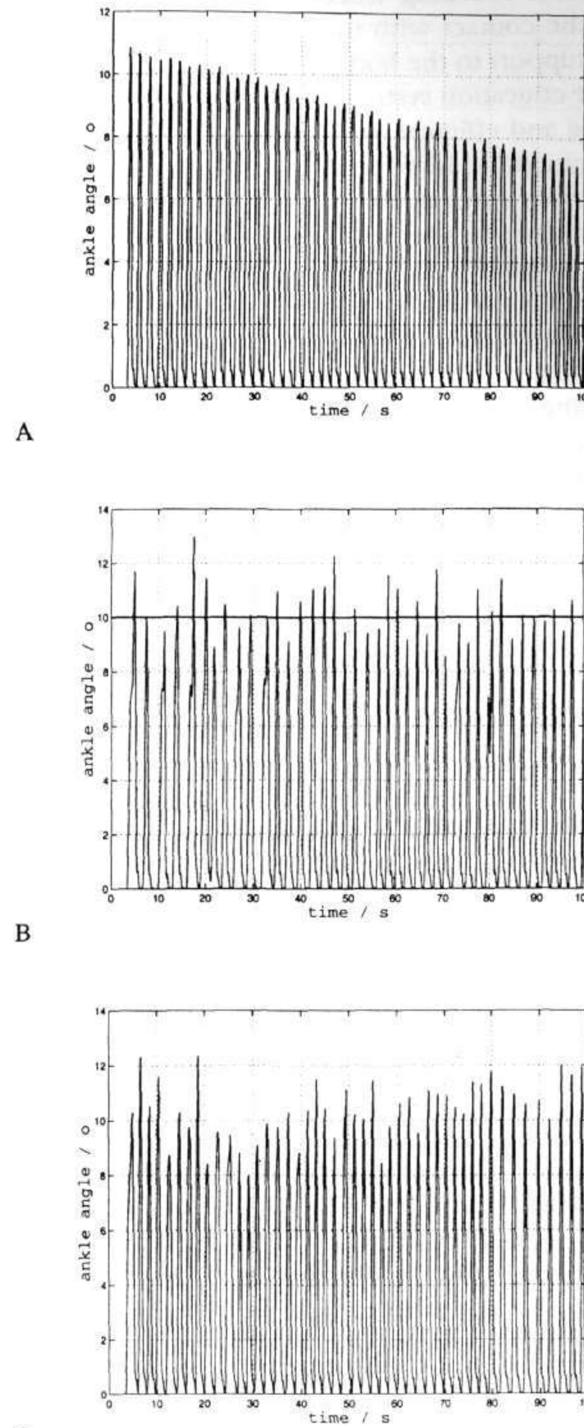

Figure 2. The ankle goniogram during simulated FES assisted walking without sensory information (a), with the use of visual and haptic feedback (b), and while using the haptic interface alone (c).

SENSORY INTEGRATED FEEDBACK SYSTEM

An incomplete SCI person's walking relies primarily upon the visual information about the position of the paralyzed extremity. Frequent inappropriate looking down to the legs is therefore necessary, resulting in rather slow walking. This is specially noticeable during the foot contact phase when the patient is checking whether the leg which comes into the contact with the ground will provide reliable support to the body. The aim of the developed FES re-education system is to provide to the patient simple and efficient information based on sensory integration(7). Information, provided in the form of sensory stimulation, is delivered to the patient's nonparalyzed upper arm in order to reward the patient for successful progression from the swing phase into the double support phase.

The laboratory version of the FES re-education system runs on a personal computer (Fig. 3). The following transducers are connected to the data acquisition module: crutch pushbuttons, knee goniometers, and foot-switches. Hand pushbuttons are built into the handles of the crutches and are used for voluntary control of a two-channel stimulator. The signals from the pushbutton are also used for recognition of the current phase of walking. When the pushbutton is pressed, the stimulation is delivered to the ipsilateral peroneal nerve resulting in flexion response, i.e., simultaneous hip and knee flexion and ankle dorsiflexion. The stimulated leg is in the swing phase of walking. The subject remains in the swing phase of walking as long as he is pressing the pushbutton. When the walking subject releases the hand pushbutton, the peroneal stimulation is discontinued and the stimulation of the knee extensors is started making the contact of the stimulated leg with the ground. The stimulation frequency was 20Hz, the pulse duration 0.3 ms, and the amplitude of stimuli up to 150 V. The knee angle was measured by the use of flexible goniometer (Biometrics Ltd, Gwent, UK). This goniometer can be easily attached to the knee joint and causes only small errors due to the skin movement. The foot-switches were attached under the heel. In most severely paralyzed incomplete SCI patients the described stimulation channels and transducers were applied bilaterally while frequently unilateral application of the FES gait re-education system was sufficient. The computer controls also an additional stimulation channel providing sensory stimulation feedback. The sensory stimulation was delivered to the patient through a pair of electrodes placed over the skin of the ipsilateral upper arm. The stimulation frequency was 50 Hz, pulse duration 0.3 ms, and the amplitude between 30 and 40 V. This reward sensory signal lasted for a predetermined time interval of 0.2 s and was generated in the beginning of the double support phase when the stimulated leg made the contact with the ground. From the control algorithm point of view, the double support phase started after the patient voluntarily released the crutch pushbutton. During this phase the patient must make a

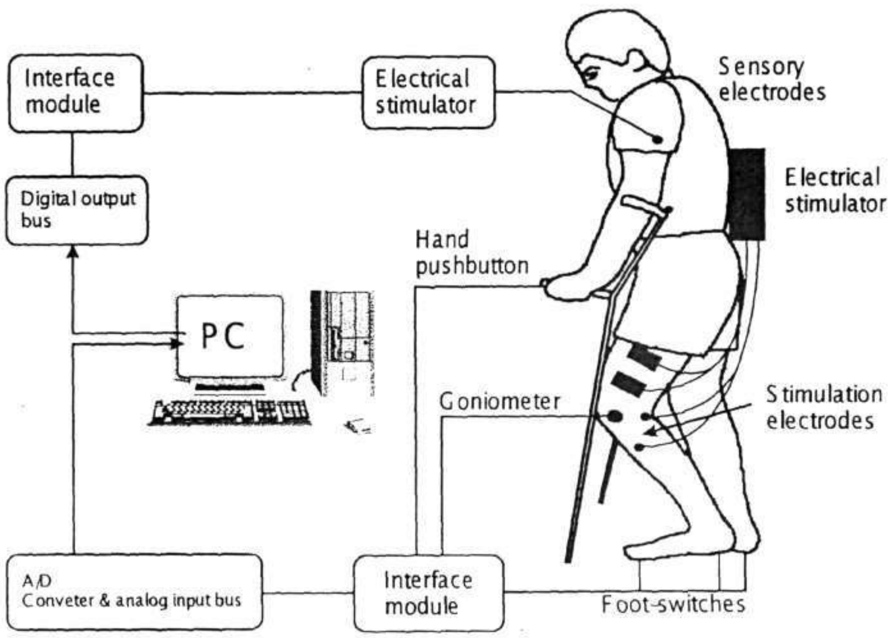

Figure 3. Sensory integrated feedback system.

contact with the ground having the leg extended. The successful foot contact was recognized when the knee angle was lesser than the maximal allowed knee flexion and the heel switch was in the ON state. In this situation the reward sensory signal was generated.

Testing of the proposed FES gait re-education system was performed in three rather severely handicapped SCI patients(7) having thoracic spinal cord lesion. Bilateral stimulation of knee extensors and peroneal nerve was needed in all three subjects. The purpose of this preliminary testing was to demonstrate that the improvements in walking occur as a consequence of applying the sensory feedback system. The gait measurements lasted for a month. During the first week the average values and variability of basic gait parameters were assessed when walking with FES but without sensory feedback. In the next three weeks the basograms were measured while training the patient to walk by the help of the described sensory feedback. The aim of the FES re-education system was to shorten the double support phase and thus increase the speed of walking. It is of utmost importance to make the double stance phase as short as possible because of fatiguing of the stimulated knee extensors. The average values and standard deviations of the double support time appertaining to the right and left leg are shown in Fig. 4. Comparing the gait data at the beginning and at the end of the investigation it can be concluded that the patient with T-12 spinal cord lesion(14 months after the accident) adopted a new and considerably faster technique for double stance phase performance.

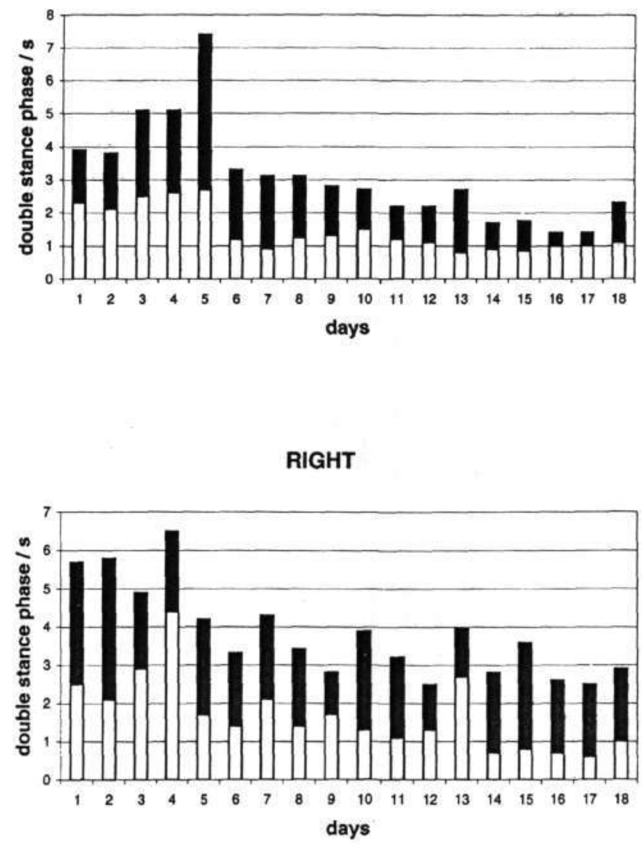

Figure 4. Influence of gait re-education on the duration of the double stance phase.

FES SYSTEM FOR PERMANENT USE

When more effects of the exercise and intensive task-dependent training are expected, the incomplete SCI patient may be a candidate for permanent application of an FES orthotic system. Simple peroneal stimulators can turn several of them into community walkers, effectively using the stimulator throuhout the day. It was our observation that the peroneal nerve stimulation was found useful in at least 10 percent of incomplete SCI patients to augment ankle dorsiflexion and knee and hip flexion in a lower limb reflex pattern.

Simple manual tests of voluntary muscle strength were performed in 31 incomplete SCI patients. The tests were carried out immediately after their arrival to the rehabilitation unit. Only the patients who were unable to walk on the day of examination were considered. The muscle groups governing the hip, knee, and ankle joint movement were evaluated. In the manual muscle tests voluntary muscle responses were estimated by six grades (0–5). The results of the muscle strength testing performed in 31 incomplete SCI patients with a central type (thoracic or cervical) of spinal cord lesion are presented in Fig. 5. It can be observed that hip and ankle antagonists were rather severely affected in most of the subjects. The strongest muscle group was knee extensors. Only rare incomplete SCI patients are candidates for application of permanent FES to their quadriceps muscles. Patients with very weak knee extensors are bound to the wheelchair. Patients with sufficently strong knee extensors are candidates for FES-assisted walking. Only one-channel electrical stimulators were given to these patients after release from the

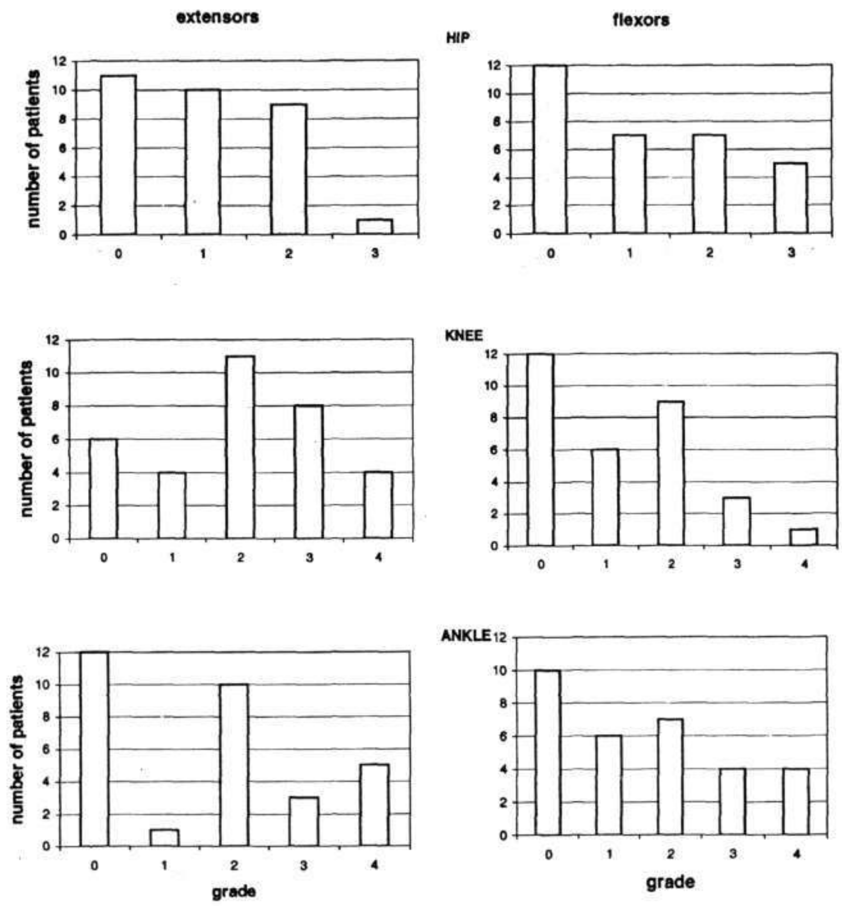

Figure 5. Distribution of the muscle strength in agonist and antagonist muscles of hip, knee, and ankle as assessed in a group of incomplete SCI persons.

rehabilitation center. One-channel FES was delivered to the peroneal nerve, resulting in a flexion response of the lower extremity. Here it also must be noted that significant nonsymmetry of the neuromuscular properties of the right and left paralyzed legs was often observed in incomplete SCI patients. In this way the peroneal stimulation was most often applied unilaterally.

It was our further observation that the incomplete SCI patients who are candidates for permanent use of the peroneal stimulator are all crutch users(2). In this respect we found it more appropriate to use the hand pushbutton built into the handles of the crutches to trigger the electrical stimulation than the more often used heel-switch. In addition, a moderate to high degree of extensor spasticity was usually observed in the lower extremities of the incomplete SCI persons. This extensor responses increased when loading the leg during standing posture or during the stance phase of walking. The extensor response is useful from the point of view of supporting the body, but is quite cumbersome during the transition from the stance into the swing phase. The patients have difficulties breaking the spontaneous extensor activity in order to be able to lift the heel and thus start the peroneal stimulation.

Interconnecting wires between the crutches and the stimulator are inconvenient in daily activities and are a frequent source of malfunctions. They hinder a patient when standing up or sitting down. The wire connection was found particularly inappropriate in situations when patients, while sitting, wish to discard the crutches. To overcome these problems, a telemetry system was developed providing reliable and interference resistant wireless control of FES-assisted walking(8). The crutch pushbutton signals are coded and transferred from the transmitter placed in the crutch to the receiver which is part of the stimulator and is firmly attached to the patient's lower leg. Another important achievement of the telemetric system is the improved appearance, since the stimulator can be hidden under the clothing of the patient.

The influence of the peroneal stimulation on gait efficiency and energy consumption was investigated. Gait performance on an incomplete SCI subject was compared to a healthy person's walking. The patient was a 48-year-old male. The level of injury, which took place four years ago, was C3-C4. The patient was using the peroneal stimulator in everyday life. Two already established gait evaluation methods were used: measurement of oxygen consumption and heart rate analysis.

The metabolic energy expenditure was calculated from the difference between oxygen consumption during walking and rest (9). The net energy physiologic cost index (EPCI) was calculated:

$$EPCI = (E_w - E_r)/v,$$

Where E_w is energy expenditure during walking, E_r is energy expenditure during rest, and v is walking speed. The heart rate was recorded by a system for physiologic measurements. Average heart rate and physiologic cost index (PCI) were calculated(10):

$$PCI = (HR_w - HR_r)/v,$$

where HR_w and HR_r denote average heart rate during walking and rest, respectively. The expired air collection and heart rate monitoring took place during the last two minutes of resting and walking. The patient walked first without any orthotic aid and afterwards by using the peroneal stimulator.

Both assessment approaches showed major differences between the gait of the normal subject and the patient. The results illustrate that the patient's gait is less energy efficient than the gait of the normal subject (Fig. 6). Performance of the patient's gait with the use of FES is much improved as compared to the gait without FES. It is evident that peroneal stimulation can significantly reduce energy consumption and improves incomplete SCI patient's gait efficiency.

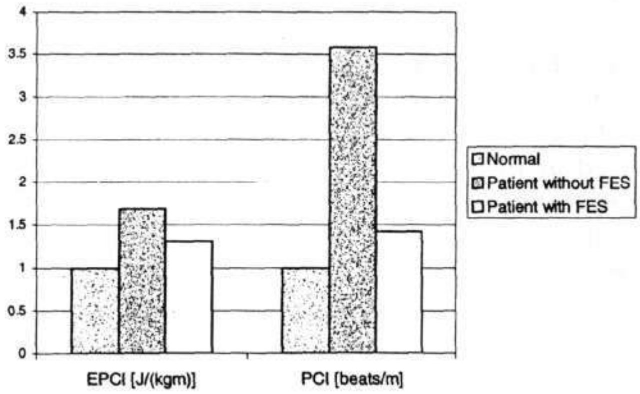

Figure 6. Energy efficiency of walking with and without FES represented by two gait assessment methods (the values are normalized to the results in healthy subject).

DISCUSSION

The use of FES for lower extremities in incomplete SCI persons can be split into two phases. First is the post-trauma phase when the stimulation is delivered to the paralyzed persons soon after the accident in the rehabilitation center. The second phase starts after the patients are released from the rehabilitation unit and when they use the electrical stimulator as community walkers during their daily activities.

The candidates for therapeutic post-trauma stimulation are all incomplete SCI patients with central lesion to the spinal cord. These are the patients with thoracic and cervical levels of SCI. The simplest application of FES is cyclical electrical stimulation delivered to a selected muscle group. It was our observation that voluntary response increased after a program of cyclical electrical stimulation in a vast majority of the incomplete SCI persons. A study should be made with a control group to distinguish the spontaneous improvements from the benefits of the FES therapeutic program. Such an investigation has not yet been performed. The reason is mainly because it is difficult to explain to the patient why he will not be included into the FES training program. Also, the cyclical electrical stimulation is a simple therapeutic program from the point of view of application of the surface electrodes and adjustment of the stimulation parameters and is quite often used by the therapists.

Electrical stimulation cannot only produce repetitive isolated movements in the paralyzed extremities. FES can provide support to the body while standing or walking and can also produce the swing phase of the extremity. The main goal of this paper was to introduce the FES-assisted gait re-education systems. The idea of these rehabilitative systems is the maximal commitment of the patient during the FES gait training. The patient has to control voluntarily the onset and intensity of the FES. At the same time the information from the paralyzed extremities is fed back to the patient. Two different groups of the FES-assisted gait re-education systems were proposed in this paper. In the first case a single gait variable was selected. Its value was analogously transmitted to the walking subject in a form of sensory feedback information. In the second case the information about several gait variables was first integrated and than delivered to the walking subject as a single command. The clinical value of both systems proposed must be further evaluated in future studies.

Many of the incomplete SCI patients recover to such an extent that they need no rehabilitative aid when released from the rehabilitation center. Nevertheless, it was our observation that at least 10 percent of the population are candidates for chronic application of an electrical stimulator. In most of incomplete SCI patients the knee extensors are strong enough to provide the necessary support during the stance phase. For chronic use these patients predominantly need only single-channel FES of the peroneal nerve resulting in the flexion response, producing the swing phase of walking. It was also observed that the candidates for permanent use of an FES walking aid are crutch users. In this way the control of FES by crutch pushbutton can overcome many problems of the heel switch triggering of the peroneal stimulator. The most important observation in this group of patients is the significant improvement in energy efficiency of walking assisted by FES orthotic device. These patients are also excellent candidates for implanted FES systems.

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Swing phase estimation in paralyzed persons walking

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Abstract. In the paper we present the sensory system aimed for FES gait reeducation of incomplete spinal cord injured persons. The proposed system consists of four accelerometers, gyro and two goniometers placed along the shank of the paralyzed leg. The data assessed are input into mathematical algorithms estimating the swing phase quality. The estimation is based on swing phase detection and signal correlation and is used to determine the cognitive feedback signal. The feedback signal is delivered to the patient as an auditory signal.

The preliminary measurements were performed in three healthy and two incomplete spinal cord injured persons with C6 and C6-7 lesion during treadmill walking. FES was manually triggered by a physiotherapist. The results have shown that the proposed multisensory system was successful in gait quality estimation. Therefore, the use of FES multisensory system and cognitive feedback could be an efficient rehabilitative approach in gait reeduction.

Keywords: Multisensory system, sensory integration, swing phase, walking, spinal cord injury, gait reeducation, functional electrical stimulation

1. Introduction

In recent years our research studies have been focused on incomplete spinal cord injured (SCI) patients. In our earlier studies we realized the necessity of functional electrical stimulation (FES) gait training in the early period after spinal cord injury [1]. The candidates were all patients with upper motor neuron lesion, in more clinical terms the patients with thoracic or cervical lesion to the spinal cord. Only a few of incomplete SCI patients were candidates for permanent use of FES, most of them used FES only during their stay in the rehabilitation center or soon after being released. In these patients peroneal nerve stimulation was found useful to provoke flexion response resulting in the swing phase of walking. Several existing systems employing peroneal nerve stimulation used sensory information to trigger FES during walking. The sensory information was usually provided by use of simple artificial sensors [2]. Data collected by a pair of miniature accelerometers were used to distinguish between the stance and swing phase. Automatic detection algorithms were used to identify the appropriate phase of walking and to control FES. On the basis of the results obtained, development of a small implantable sensor-stimulator device was proposed. Dai et al. [3] proposed an application of tilt sensors in FES and used previously commercially available functional electrical stimulator. Williamson and Andrews [4] presented a gait event detection using three uniaxial accelerometers mounted below the knee of the patient. The gait

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event detection was based on rule based detectors and adaptive logic networks to distinguish between stance and swing phases during the gait cycle and detect the transitions between five gait phases during walking. Kostov et al. [5] used the adaptive logic networks to control the functional electrical stimulator for foot drop correction. The sensory signals were assessed from ENG and a heel switch. All these sophisticated techniques for gait detection were used to trigger the electrical stimulation, but in practice most of the systems in the rehabilitation environment still use the foot switch [6].

The aim of a FES rehabilitative system for re-education of walking [7] is not only to deliver electrical stimulation to the paralyzed muscles, but also to assess the sensory information from the paralyzed limb. The sensory information is fed back to the patient and not to the stimulator control unit. The FES rehabilitation systems for re-education of walking are intended to be used with incomplete SCI persons soon after the accident or onset of disease. These systems are to be used within the rehabilitation centers and applied by therapists. Surface electrical stimulation is therefore appropriate. We are developing two separate systems for training of proper movements during swing and stance phase. The adequate approach is selected according to the patient's gait deficits. In this paper we are describing the swing phase quality estimation. The gait re-education system for swing phase detection and swing quality estimation is based on multisensor use, simple feedback signal, which is fed back to the patient, and FES. The feedback signal can be delivered to the patient through vibrotactile or electrical stimulation or by the help of an auditory signal. It represents the successfulness or unsuccessfulness of performing the swing phase. In the envisioned gait reeducation system the patient has the possibility to control voluntarily the intensity of FES by a manual control lever [8] in order to improve swinging of the paralyzed lower extremity.

The preliminary measurements, described in the presented paper, are aimed to determine what is an appropriate swinging of the lower extremity. In incomplete SCI, when usually one side is affected considerably more than the other, we want to achieve symmetry of the right and left leg swing. Therefore, the gait reeducation is focused on making the swinging movement of the affected leg similar to the less-affected leg.

2. Methods

2.1. Hardware description

We used the sensory system [9] employing two goniometers (Penny and Gilles), single-axial gyroscope (Murata ENC 03JA) and two pairs of single-axial accelerometers (ACCESS). Accelerometers were mounted on a small aluminum plate and mounted perpendicularly in pairs. The gyroscope was mounted on a board with analog bandpass filter and amplifier and placed in the middle of the plate (Fig. 1). The plate (dimensions 192×42 mm) was attached to the shank of the patient by the help of velcro straps. All sensory signals were low-pass filtered, using the third order Butterworth filter and 10 Hz cut-off frequency, besides gyroscope signal which was first high-pass filtered (0.05 Hz) in order to avoid the temperature drift. The personal computer (PC) with Pentium® III 500Mhz and Burr-Brown acquisition board were used to assess the data. The data were sampled at 100Hz with the resolution of 12 bits. A computer controlled four channel electrical stimulator developed in our laboratory was used. Single channel surface peroneal nerve stimulation was delivered to the patient. The stimulation frequency was set to 20 Hz while pulse duration and stimulation amplitude were adjustable. The stimulator was controlled by PC via RS232 and was triggered by manual switch.

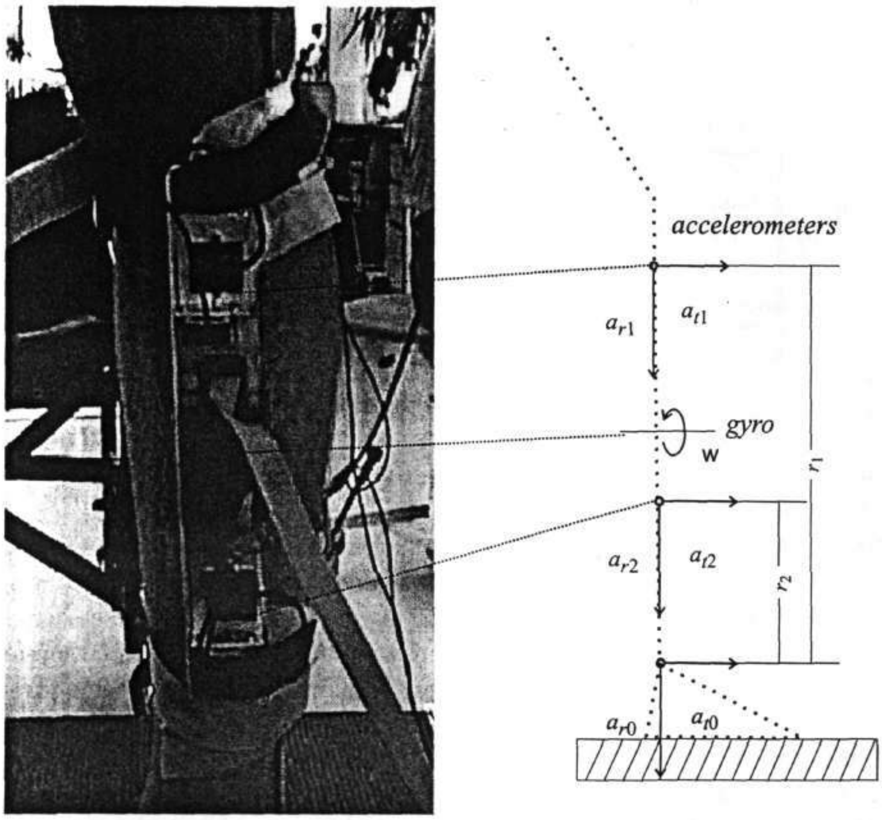

Fig. 1. Multisensor device attached to the shank. It consists of two pairs of single axial accelerometers and a gyroscope. Two goniometers are attached to the lower extremity to assess the ankle and knee joint angles. On the left the photograph of the frontal plane is shown, while the schematic presentation on the right shows the parameters measured in the sagital plane.

2.2. *Swing phase estimation algorithm*

Multisensor system was assigned several functional tasks. The gyroscope signal was used for swing phase detection, while the other sensors took part in swing quality estimation. The multisensor system was first placed on the shank of the less-affected extremity to assess the ankle joint acceleration during treadmill walking. The time-course of the acceleration assessed during the swing phase was used as a reference a_{ref} for the more affected extremity during the gait reeducation. The acceleration signals were measured at two points on the shank. At both points radial (a_{r1}, a_{r2}) and tangential (a_{t1}, a_{t2}) acceleration were measured. By use of a simple Eq. (1) we get the acceleration in the ankle joint [2]:

$$\begin{bmatrix} a_{t0} \\ a_{r0} \end{bmatrix} = \frac{1}{r_1 - r_2} \cdot \left(r_1 \cdot \begin{bmatrix} a_{t2} \\ a_{r2} \end{bmatrix} - r_2 \cdot \begin{bmatrix} a_{t1} \\ a_{r1} \end{bmatrix} \right) \tag{1}$$

In the Eq. (1) r_1 and r_2 present the distances from ankle joint to the place where particular pair of accelerometers is attached (Fig. 1).

Two algorithms are related to the signals of the multisensory system, swing quality estimation algorithm and swing phase detection algorithm. Inputs to the swing quality estimation algorithm are a pair of radial (a_{r0}) and tangential (a_{t0}) acceleration, ankle and knee joint angles (θ_1) and an output of the swing phase detection algorithm. The aim of the swing quality estimation algorithm is to provide reliable swing quality coefficient which determines the cognitive feedback signal.

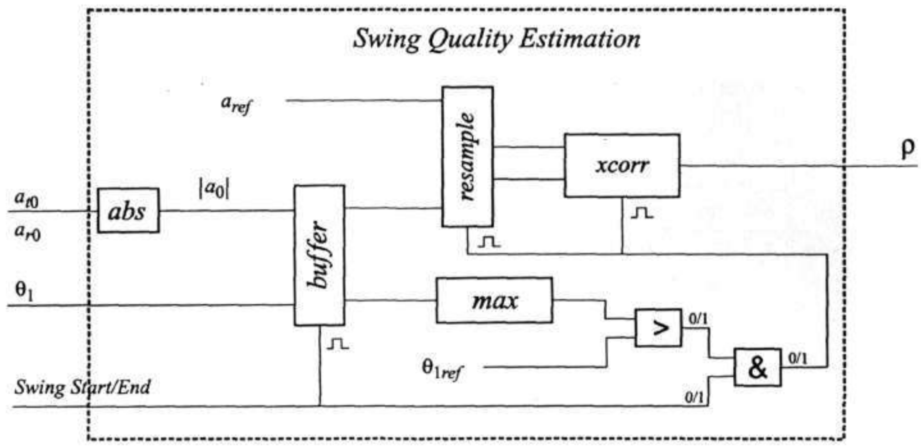

Fig. 2. Swing phase estimation diagram.

Swing phase detection algorithm represents the important part of the system, since it is our aim to estimate the time-course of the signals only during the swing phase. When a lower extremity enters into the swing phase, there occurs a significant change in the angular velocity of the shank. The angular velocity changes the sign from negative to positive crossing the zero what is in our system detected as the start of the swing phase. The end of the swing phase is detected with the same presumption. In a case of unstable knee [10] a false swing phase can be detected during the stance phase. Such a gait event is recorded, but not considered in further processing. The swing phase detection algorithm is the basic part of the swing reeducation algorithm and is input into every other algorithm.

In swing quality estimation (Fig. 2) the swing phase detection algorithm provides the start and the end of signal processing. The swing phase detection signal is set to 1 during the detected swing phase and thus enables the swing quality estimation algorithm. During the swing phase the absolute value of the ankle joint acceleration ($|a_0|$) and knee joint angle (θ_1) time-courses are recorded and stored in a buffer. At the moment, when the end of the swing phase is detected, the data assessment is completed. The algorithm examines the measured knee joint angle. If the maximum value of the knee joint angle exceeds θ_{1ref}, then a swing phase estimation algorithm is calculated, otherwise the swing phase estimation is rejected and the swing is marked as poor. The θ_{1ref} is defined on the basis of previous measurements, patients deficits and therapist requirements. Usually it was expressed in percent of the less-affected leg (80–90%) maximal knee flexion. In most cases the swing phase duration varies from the duration of the desired swing phase. Consequently, the number of samples in each time-course is different. Before making a comparison between the acceleration samples of both legs, a resampling is needed. The absolute value of the recorded ankle joint acceleration [9] is afterwards correlated with the reference acceleration time-course. The reference a_{ref} has been previously recorded on the less-affected extremity and represents the desired swing phase. The assessment cycle is repeated during every swing phase. The correlation is calculated on a basis of the following Eq. (2):

$$\varphi_{m,r}(t,\tau) = \frac{1}{n}\sum_{k=1}^{n} a_m(t)\cdot a_r(t+\tau) = E[a_m(t)\cdot a_r(t+\tau)] \tag{2}$$

where τ means a time delay and n is a number of samples included into computation. In the Eq. (2) a_m is the measured acceleration, while a_r is the reference acceleration time-course. Afterwards the correlation coefficient is calculated:

$$\rho_{m,r} = \frac{E[(a_m(t) - m_m(t)) \cdot (a_r(t+\tau) - m_r(t+\tau))]}{\sqrt{E[(a_m(t) - m_m(t))^2 \cdot (a_r(t+\tau) - m_r(t+\tau))^2]}} \tag{3}$$

where m represents the mean of the signal. When the calculated coefficient $\rho_{m,r}$ is close to 0, there is no correlation between the signals and closer to 1 it gets, more signal resemblance exists. The correlation coefficient presents the criterion of the swing phase appropriateness (0 – poor, 1 – perfect).

2.3. Cognitive feedback

The cognitive feedback is defined on the basis of the correlation coefficient $\rho_{m,r}$. The feedback signal is divided into three discrete levels. In case of a low correlation coefficient the swing was deemed as poor. When the coefficient reached 0.2, swinging of the leg was considered sufficient and above 0.6 good. These criteria should be adapted to individual patient. The coefficient limits are set according to the patient's deficits and the requirements of the therapist. The feedback signal in this preliminary investigation was delivered as auditory cognitive feedback and was provided by PC as a sound of three different frequencies for sufficient, poor, sufficient and good.

3. Results

In the preliminary investigation three healthy subjects and two incomplete SCI patients were involved. All three healthy subjects and the patients with C6-7 and C6 lesion to the spinal cord, were walking on level terrain in the laboratory. The patient LG with C6 lesion (9 years after accident) was not an FES user. He was using crutches during walking. The other patient BM had a C6-7 lesion (18 years after accident) and was trained to use FES. FES assisted training of walking was performed on a treadmill. The treadmill speed was first set to 0.7 m/s and later decreased to 0.5 m/s. During walking we used the described multisensor measurement system [9] together with the sensory feedback algorithm running on PC, using Matlab® software. The proposed hardware and software were first tested in healthy subjects. Figure 3 presents the signals assessed from gyro and goniometers together with the calculated absolute value of the ankle joint acceleration. The gyro waves correspond to the swing phase of walking. It is obvious that the acceleration during the stance phase is equal to the gravitational acceleration and that the maximum knee angle is reached during the the swing phase. The ankle goniogram shows dorsiflexion throughout the swing phase.

In Fig. 3 there was no error in detection and all swing phases were satisfactory. This was expected since walking of a healthy person is quite regular. In figure the swing phase detection is presented by rectangular pulses. During swing phase the signal is set to the value 100 otherwise it is 0. The swing phase signal corresponds to zero crossing of the gyro signal. The lower figure presents the correlation coefficient which was almost all the time above 0.6, representing thus a perfect swing.

The patient LG was walking on level terrain supported by crutches (Fig. 4). His walking pattern was irregular during the stance phase. Often patients with spinal cord injury avoid knee flexion in loading response of the stance phase because of quadriceps weakness. This can occur when most of the body weight is carried through the arm support provided by the crutches. Consequently, the gyro signal detected this event as swing phase. Furthermore, the algorithm marked it as a false detection and provided a negative feedback signal. It is shown in the lower chart (Fig. 4) as a negative value -1. As

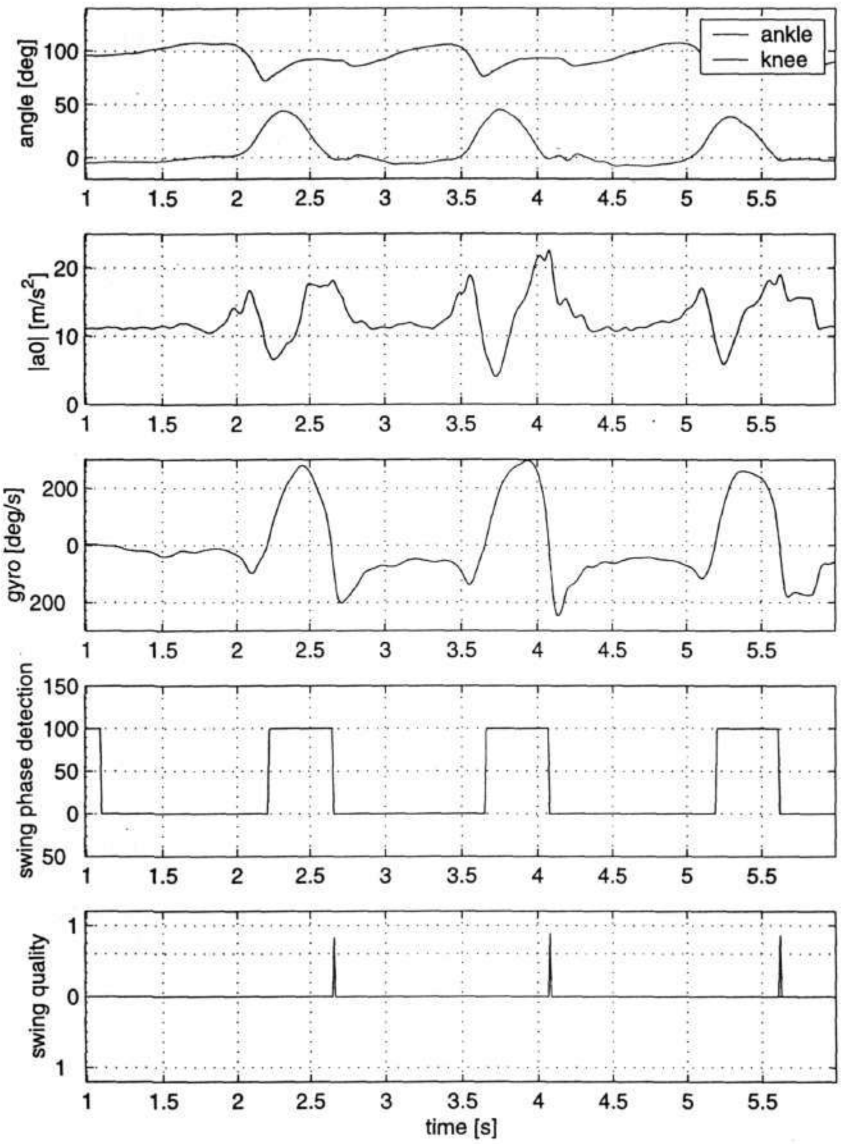

Fig. 3. Time-course of gyroscope, absolute value of ankle joint acceleration and ankle and knee joint angles during level walking of a healthy person. The two charts below present the swing phase detection and the swing quality coefficient with a threshold for good swing.

the patient was a trained crutch walker, the quality of his swing phase was quite good except the last swing was extremely poor. The unstable foot [10] contact caused a swing phase extension which poorly correlated with the reference acceleration.

Patient BM was not a regular FES user. In the rehabilitation center he was using a single channel MicroFES peroneal brace together with crutches, (Institut Jozef Stefan, Ljubljana). He was training FES walking for two months. We measured his walking on a treadmill with and without FES. The patient was asked to walk continuously for 60 s. After every trial he took a rest of 5 minutes. During FES assisted walking the physiotherapist was manually triggering the surface peroneal nerve stimulator according to the provided cognitive feedback. Here we present only the measurement with FES, since

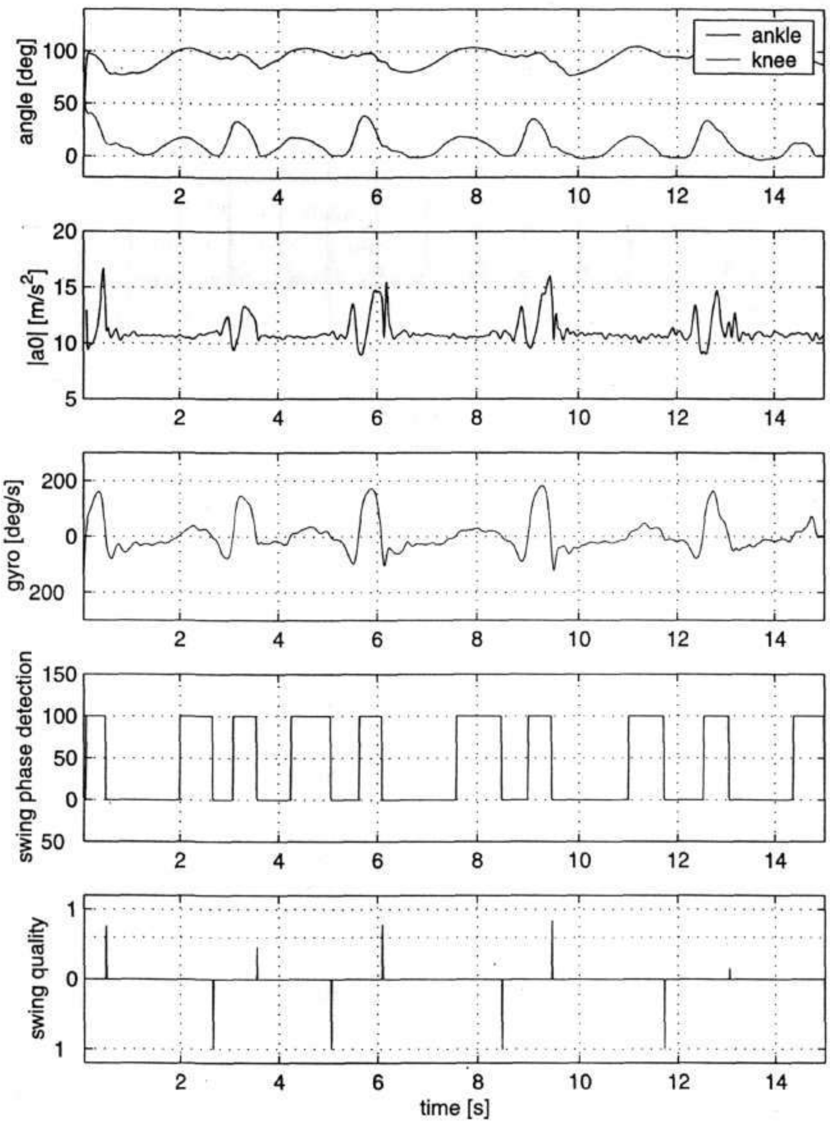

Fig. 4. Time-courses of ankle and knee joint angles, absolute value of ankle joint acceleration and gyroscope during level walking of patient LG using crutches are presented. Swing phase detection and swing quality are presented in two lower charts. False detections occurred in stance phase and were successfully eliminated. These errors are presented in the lowest chart by the negative value -1 of the swing quality indicator.

swing quality can be improved by selecting adequate start, duration, and intensity of train of electrical stimuli delivered to the peroneal nerve. The first trace of Fig. 5 presents the gyro signal which was used as the swing phase detector. The absolute value of the ankle joint acceleration $|a0|$ during the swing phase was selected as the main criterion for the swing quality estimation. During the stance phase it was equal to the gravitational acceleration, while during the swing phase it comprised noticeable waves and spikes which were used for determination of correlation coefficient. The joint angles in ankle and knee are also presented in Fig. 5. The knee goniogram shows obvious difference with regard to walking of a healthy person. The presence of FES is also shown in Fig. 5. In the last trace the correlation coefficient is presented. The swing quality is quite low, since patient BM was using FES for a short time only.

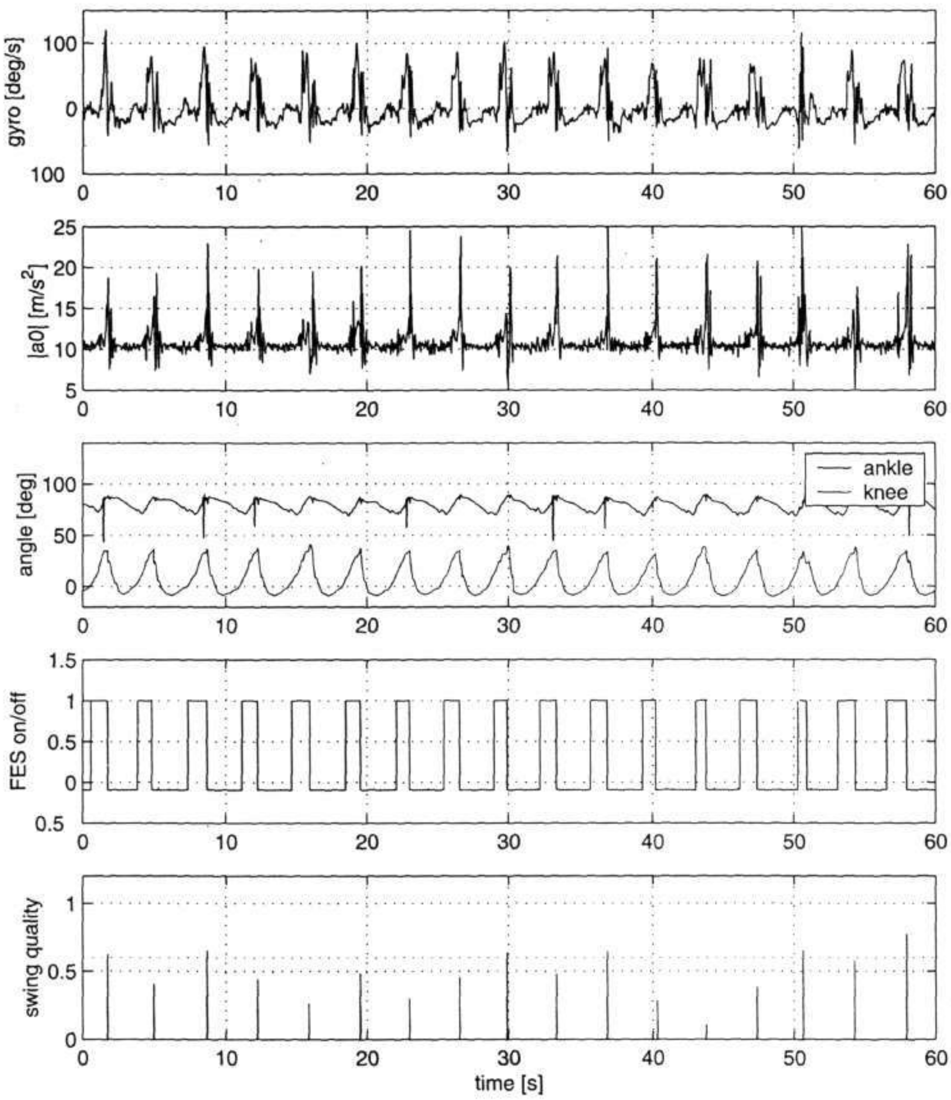

Fig. 5. Sensory signals and correlation coefficient during treadmill walking of patient BM. The cognitive audio feedback was provided according to the swing quality. FES was manually triggered by the physiotherapist.

The swing quality recorded in SCI person differs significantly from that assessed in healthy person. The less-affected extremity swing of the incomplete SCI patient was used as a reference. Cognitive audio feedback was based on correlation coefficient and served as a warning to the physiotherapist who was manually triggering FES by a pushbutton.

4. Conclusions

On the basis of the results obtained we can describe the swing quality estimation algorithm as a successful attempt of multisensor use in the rehabilitation environment. There were several difficulties encountered in the detection of the swing phase of a SCI person. All these false detections were filtered out so that there was no influence on the feedback signal at the end of the swing phase. The presented preliminary measurements have shown the feasibility of the multisensor use in the gait reeducation. The proposed algorithm can help physiotherapists in the early stage of rehabilitation of SCI patients. In these preliminary experiments the physiotherapist triggered FES according to the auditory feedback. Further development will make possible the FES triggering by the patient. In future measurements we shall have a pushbutton mounted on the treadmill supporting frame to involve the patient directly into the rehabilitation process. Further development will make the patient possible to control the stimulation amplitude by replacing the pushbutton by a special trigger [8]. In this way the patient will be even more actively involved in the gait reeducation what is of utmost importance for the successfulness of the rehabilitative approach.

Acknowledgments

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Dodatek D

FES Gait Re-education: The Swing Phase Estimation

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■ **ABSTRACT**

This paper presents the use of multiple sensors for walking assessment and provision of cognitive feedback during early re-education of incomplete spinal cord injured (SCI) humans. The paper is focused on the swing phase estimation as an important part of the Functional Electrical Stimulation (FES) gait re-education system for incomplete spinal cord injured persons. The proposed sensory system comprises four accelerometers, one gyro placed at the shank of the paretic leg, and two goniometers placed at the knee and ankle joints, respectively. The data from the sensors are input in the mathematical algorithm applied for swing quality estimation. The output from the algorithm is a numerical value. The calculated output is divided into three levels, each defining the swing quality in terms of good, sufficient, and poor. This information is provided to the patient as an auditory signal. The patient is taught to maximize his efforts to improve the quality of walking, that is, to move the more affected leg in a way that will generate the auditory output corresponding to the level "good".

The preliminary measurements were performed in healthy subjects walking on even terrain and in an incomplete SCI person with C6 lesion during walking on the treadmill. FES in the latter case was triggered manually by a physiotherapist. The results showed that the timing of FES triggering played an important role in sensory-supported FES-assisted walking, that is, the auditory feedback was also a cue to the therapist controlling the FES. The swing quality estimation enabled patients to voluntarily improve their walking, consequently the intensity of FES assistance was decreased. This suggests that the use of an FES multisensor system for cognitive feedback is efficient rehabilitative method in early stage of rehabilitation of walking. ■

KEY WORDS: functional electrical stimulation, gait, incomplete spinal cord injury, multisensor system.

INTRODUCTION

In recent years our research studies have focused on incomplete spinal cord injured (SCI) patients. In our earlier studies we realized the necessity of FES gait training in the early period after spinal cord injury (1). The candidates were patients with upper motor neuron lesions, that is, patients with thoracic or cervical lesions to the spinal cord. Since they were incomplete SCI patients, their movements were disturbed or they were not able to dorsiflex the ankle. Other motor deficits included insufficient knee and hip flexion. Only a few incomplete SCI patients were found to be candidates for permanent FES, and most of them used FES only for a period of about one year after being released from the rehabilitation center (2). In these patients peroneal nerve stimulation was found

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useful to provoke flexion response resulting in the swing phase of walking.

Several rehabilitative systems employing peroneal nerve stimulation used sensory information in order to trigger FES during walking. The sensory information was usually provided by simple artificial sensors. Data collected by a pair of miniature accelerometers were used to distinguish between the stance and swing phase (3). Automatic detection algorithms were used to identify the appropriate phase of walking and to control FES. On the basis of the results obtained, development of a small implantable sensor-stimulator rehabilitation device was proposed.

The aim of a FES rehabilitative system for re-education of walking is not only to deliver electrical stimulation to the paralyzed muscles, but also to assess the sensory information from the paralyzed limb. The sensory information is provided to the patient as well as to the stimulator control unit. The FES rehabilitation systems for re-education of walking are intended to be used soon after the accident or onset of disease (2). These systems are to be used within rehabilitation centers and applied by therapists. Surface electrical stimulation is therefore appropriate. We are developing two separate systems intended for swing or stance phase re-education. The adequate approach should be selected according to the patient's gait deficits. In this paper we are proposing swing phase quality estimation. In incomplete SCI individuals, where usually only one side is affected, we aim to achieve symmetry of right and left leg swinging. Therefore, the gait re-education must be focused on making the swing phase of the affected leg similar to the nonaffected swing phase. We are proposing a FES gait re-education system based on a multisensor approach, where a simple feedback signal is delivered to the patient. The feedback signal can be delivered to the patient through vibrotactile or electrical stimulation or by use of a small earphone. The cognitive feedback (CF) signal represents the degree of success of performing the swing phase. The patient can also voluntarily control the amplitude of FES by a control lever (4) in the handle of the crutch in order to improve swinging of the paralyzed lower extremity. The proposed CF could help the patient and the physiotherapist to accompany the estimated swing quality. This allows the patient to make better voluntary movement within his capabilities and the physiotherapist to hit the right instant of the FES triggering. The physiotherapist could also focus on the duration of stimuli and as will be presented had a great impact on the swing quality coefficient. In previous work (2) we emphasized the FES triggering done by the patient himself, in this study we present the task of the physiotherapist.

MATERIALS AND METHODS

The method that we propose uses the swing phase of the less affected leg as a reference for the more affected leg during the re-education of walking. The goal is to achieve symmetrical swing phases in SCI humans with incomplete lesions. The reference measurement starts from the less affected side, and the results are considered as a reference swing phase. During training or preliminary measurement the recorded reference was compared to the actual swing movement. The assessment was done by a multisensor device (5,6) consisting of two pairs of single axial accelerometers (ACCESS, Les Charbonniers, Switzerland), a single axial gyroscope (Murata ENC 03JA, Japan) placed at the shank and two goniometers (Biometrics Ltd., UK) placed at the knee and ankle joints, respectively, using 100-Hz sampling frequency. All four accelerometers were used to determine the radial and tangential ankle joint acceleration (5,6). The ankle joint acceleration was determined by using two pairs of mounted accelerometers. A tangentially mounted pair of accelerometers was used to determine the tangential component of the ankle joint acceleration and the radial component was determined the same way. The absolute value of the ankle joint acceleration $|a_0|$ was calculated from tangential and radial component and applied in all calculations. Most of the software was implemented on personal computer (PC) using MathWorks Matlab, Simulink software (MathWorks, Natick, MA).

The training of walking took place on a treadmill. The first stage of every measurement or training was to determine the reference swing phase. It was possible to select a swing phase pattern typical for an able-bodied person, but then the goal seemed to become unreachable. We suggest that at least in the beginning the pattern of the less affected leg should be used as a reference for symmetry. The multisensor device was attached to the less affected leg and assessed data during several walking cycles on the treadmill. The time-course of the assessed ankle joint acceleration in one swing phase was then used as a reference for training or only comparison with the more affected leg during walking on the treadmill.

In order to distinguish between the swing and the stance phase during walking, detection of the swing phase onset detection was needed. The algorithm was based on data assessed by a gyroscope. When a lower extremity came into the swing phase, then a significant change of the angular velocity of the shank would occur. This change was detected, and assigned as the beginning of the swing phase. The end of the swing phase was detected by using the same presumption. During the swing phase the ankle joint acceleration (a_0) and knee joint angle time-courses were recorded and stored into a buffer.

The swing phase estimation algorithm is based on two inputs, a flag and two input references. When the swing phase takes place, a flag is set to a positive value enabling the two inputs, which store the data from accelerometers and knee joint goniometer into a buffer. At the end of the swing phase the flag is set to zero and data assessment is thus finished. One of the input references is a user-set demand for minimum knee flexion during swing phase. Therefore the maximum value of the buffered knee joint angle time-course is searched. The maximum value presents the knee joint flexion in a preswing phase. When the user demand is fulfilled the swing phase estimation algorithm proceeds, otherwise the session is finished, and the swing is marked as poor with a corresponding zero output. The second reference is the stored prerecorded acceleration time-course. Since the goal is to provide qualitative swing estimation, a correlation with the actual assessed ankle joint acceleration is calculated at the end of the swing phase:

$$\begin{aligned}\varphi_{meas,ref}(kT,\tau) &= \frac{1}{n}\sum_{k=1}^{n} a_{0meas}(kT)a_{0ref}(kT+\tau) \\ &= \mathrm{E}[a_{0meas}(kT)a_{0ref}(kT+\tau)]\end{aligned} \tag{1}$$

The next phase is calculation of the correlation coefficient:

$$\rho_{meas,ref} = \frac{\mathrm{E}[(a_{0meas}((kT) - m_{meas}(kT))\cdot(a_{0ref}(kT+\tau) - m_{ref}(kT+\tau))]}{\sqrt{\mathrm{E}[((a_{0meas}(kT) - m_{meas}(kT))^2\cdot(a_{0ref}(kT+\tau) - m_{ref}(kT+\tau))^2]}} \tag{2}$$

where m represents the mean of the signal, n number of samples, a_{0ref} the stored reference acceleration and a_{0meas} the assessed ankle joint acceleration. This algorithm is performed at every swing phase end, except when user demands for knee flexion have not been not fulfilled. The calculated correlation coefficient presents the signal similarity. At zero output or low correlation coefficient we have no matching between the actual swing and the desired swing. The higher is the coefficient, the better the similarity and the actual swing quality is closer to the desired.

On the basis of the correlation coefficient we defined the cognitive feedback. The problem arose how to present the feedback information to the walking SCI subject. According to previous work (7) the cognitive feedback should be simple and easy to perceive. The patient should focus on the walking, not on the feedback. Therefore, we defined only three levels for audiocognitive feedback. In case of insufficient knee flexion or low correlation coefficient the swing is deemed poor. When the coefficient is higher than 0.2, then a swing phase is considered sufficient, finally when the coefficient is above the value 0.6 the walking is rated as good. These criteria can be set manually for each individual patient. The coefficient limits are set according to the patient's deficits and therapeutic requirements. In this study we selected the auditory cognitive feedback. Personal computers provided a sound at three different frequencies that were delivered to the patient through loudspeakers.

In a case of incomplete SCI, FES assistance is needed to perform the expected quality of the lower extremity movement. Here a single channel surface peroneal stimulation is used, triggered by a physiotherapist pressing the pushbutton. Since the physiotherapist could follow the CF and is able to visually accompany the patient walking, there are no specific mapping regarding the FES triggering. We rely upon physiotherapists' experience and CF, which supplemented the visual feedback and returned valuable estimation.

RESULTS

Three healthy subjects and an incomplete SCI patient were subjected to preliminary measurements. The developed algorithms for swing phase

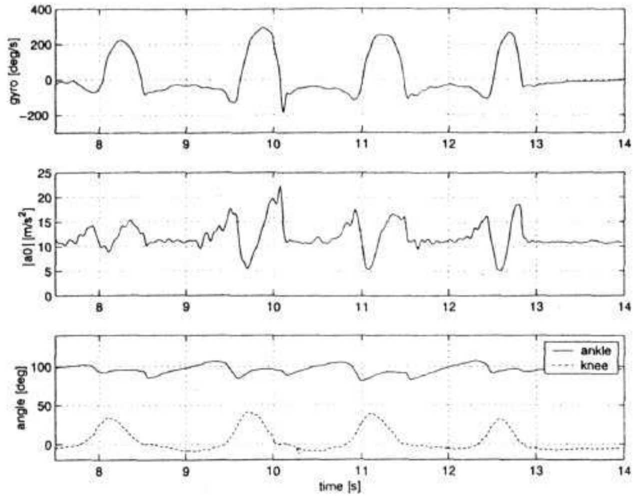

Figure 1. Time-course of gyroscope signal, absolute value of ankle joint acceleration and ankle, and knee joint goniograms during walking of a healthy person on even terrain.

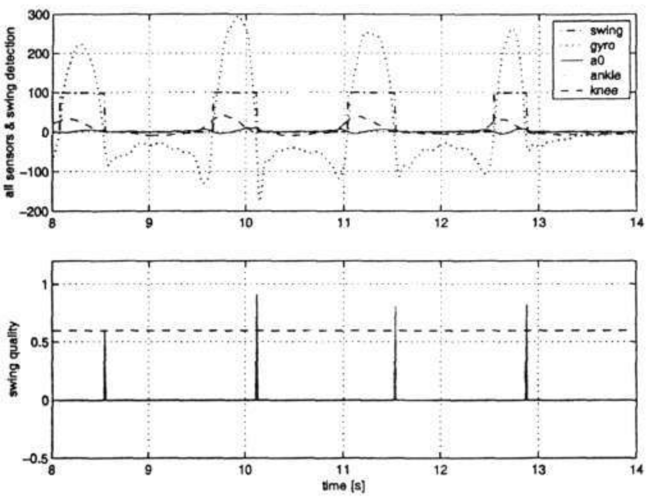

Figure 2. The output of the swing detection algorithm is presented together with signals assessed from gyro and goniometers to emphasize the duration of the swing phase. The swing phase denotes the moment of signal assessment. The appertaining value of the correlation coefficient during walking of a healthy person on even terrain is presented in the lower figure.

detection and swing phase estimation were first tested in the healthy subjects. The swing and stance phase was easily recognized in their walking (Figs. 1 and 2). Figure 1 presents the signals assessed from the gyroscope, the absolute ankle joint acceleration $|a_0|$, and the goniograms of both joints, knee and ankle, while Figure 2 shows all of the signals assessed together with the output of the swing phase detection algorithm. The chart below presents the output of the swing phase estimation algorithm. It represents the correlation coefficient ρ that was calculated by Equations 1 and 2 where a_{0ref} was assessed in preliminary measurement from the right leg and a_{0meas} from the left affected leg. As expected, the swing quality was high since walking of the healthy subject was almost symmetrical.

The second measurements was carried out at the rehabilitation center. The patient with C6 lesion to the spinal cord walked on a treadmill. The goal was to test the suitability of walking re-education based on gait symmetry (8) and at the same time to examine the adequacy of the auditory feedback. The treadmill speed was first set to 0.7 m/s and later decreased to 0.5 m/s. During the assessment of the reference acceleration signal from the less affected leg, we did not use any FES. In further walking trials the peroneal surface stimulation was triggered by a physiotherapist using the hand pushbutton. Here, we present only the measurement results from the walking with FES, since those are relevant for swing quality estimation. Figure 3 presents the signals assessed from the gyroscope, absolute ankle joint acceleration $|a_0|$, and both goniograms, together with the FES sequence, and the correlation coefficient value representing the swing phase quality.

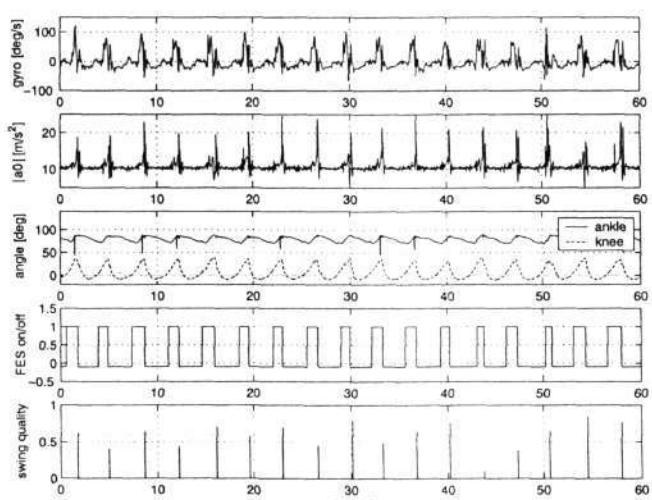

Figure 3. Time-course of gyroscope, absolute value of ankle joint acceleration, and ankle and knee joint angles during FES-assisted walking of C6 patient walking on treadmill. The output of the swing detection algorithm and the appertaining value of the correlation coefficient while using FES is also presented. FES was triggered by the physiotherapist pushing the pushbutton.

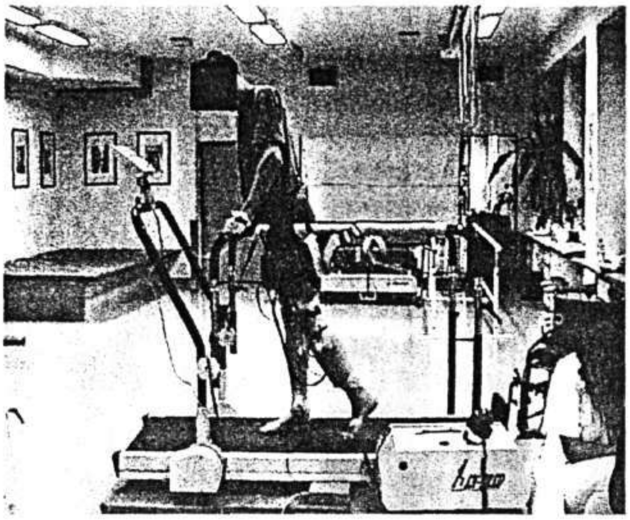

Figure 4. The incomplete C6 SCI patient during walking on a treadmill. The more severely affected left lower extremity is in the initial swing phase.

There was a significant difference between healthy person's and incomplete SCI patient's walking (Fig. 4). The heel contact can be obviously noted as peaks in the acceleration signal in Figure 3. In general this had no influence on the results, since we used the data from a less affected extremity of the incomplete SCI patient as a reference signal rather than use the prerecorded ankle joint acceleration from a healthy extremity. The lower chart in Figure 3 presents the correlation coefficients. Some double signals can be noticed in the record. They were a consequence of misdetection of the end of the swing phase, as the patient's walking was very irregular. The algorithm (5) was able to diminish the misdetection, so the CF was not impeded and was provided only at the end of the swing.

DISCUSSION

The shortening of rehabilitation of humans after SCI is extremely important. Many recent studies suggest that this can be achieved by combining voluntary and externally augmented movement exercises, that is, by involving the patient into the process of re-education. FES gait re-education is an excellent candidate for speeding up the rehabilitation. It allows a patient to relearn the phases of walking by using information gained by a multisensor system in form of a qualitative information about walking. This is allowed by adding the cognitive feedback. The patient is aware of his well or poorly performed swing. When the swing is poor the patient can voluntarily try to perform the better swing. At the same time cognitive feedback helps the physiotherapist to find the appropriate instant of triggering and the duration of FES. After the physiotherapist has become aware of the poor swing, she can extend or shorten the duration and appropriately trigger the FES earlier or later, depending on swing quality results, subjective visual impression, and experience. Without qualitative estimation we were not able to assure repeatability during FES training. The multisensor system and cognitive feedback resolve the problem of repeatability and applicability in clinical environment.

Results strongly suggest that the timing of FES triggering plays the important role in swing phase. All of the patients' efforts to improve the swing could be devastated by an improper triggering moment. Therefore auditory feedback is effective also to a therapist who is controlling the FES system. In addition, in order to improve repeatability, the physiotherapists' task could be replaced by computers. We propose the modification of the present system by adding automatic FES triggering and control of the stimulation intensity. In parallel, the patient's swing improvement could consequently decrease the stimulation intensity if and when required.

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Dodatek E

J Rehabil Med 2003; 35: 1–5

DEVELOPMENT OF A GAIT RE-EDUCATION SYSTEM IN INCOMPLETE SPINAL CORD INJURY

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Objective: **The aim of the paper is to present the development of a system for swing phase restoration in patients with incomplete spinal cord injury.** ***Methods:*** **The functional electrical stimulation based gait re-education system comprises a sensory system, a system providing cognitive feedback and a motor augmentation system facilitating and correcting the movement of the swinging extremity. Mathematical algorithms estimate swing quality and classify the swing phase of walking into 3 levels, termed cognitive feedback, which is provided to the patient as an auditory signal. A single channel peroneal functional electrical stimulation was applied as a motor augmentation system to provide the patient with the motor assistance required. The important novelty of the proposed system is that motor assistance is provided only at the level that enables the patient to perform a good swing.** ***Results:*** **The developed system was tested in a patient with incomplete spinal cord injury, with C4–5 lesion, whilst walking on a treadmill. The results show that the automated sensory driven functional electrical stimulation augmentation system, providing only the minimal assistance required based on the subject's performance, is a viable approach that successfully releases a therapist from the task of delivering properly timed stimulation of adequate intensity in assisting the swing phase of walking.**

Key words: central nervous system, electrical stimulation, sensory system, gait re-education, treadmill walking.

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INTRODUCTION

Over the last few decades there has been a trend of increasing numbers of incomplete spinal cord injuries (SCI), where one leg is usually considerably more affected than the other. One of the main goals of rehabilitation in paretic individuals is improvement of walking ability. Within the last 10 years novel methods of neurological rehabilitation have been developed which take into the account the plasticity of the injured central nervous system (CNS). A general methodological framework has been developed which emphasizes that sensory and motor augmentation should be set at the level that enables the execution of the practised task, thus facilitating the reorganization of the injured CNS, leading to functional improvement. When the functional task is walking, treadmill training with partial body weight support (BWS) is the method that has been developed over the years and is used today in most rehabilitation environments. Several large-scale controlled clinical studies (10) have proven the efficacy of the approach in acute as well as chronic conditions. Unfortunately, the major deficiency of the proposed method is that it requires a therapist to assist the walking subject. The therapist sits alongside the treadmill assisting the stepping movements of the paretic leg, which involves the therapist in strenuous work under ergonomically unfavourable conditions. Additionally, the assistance provided by the therapist is variable. This variation is even more pronounced with greater training time and increasing fatigue of both the therapist and the walking subject.

Different robotic systems have been developed (GaitTrainer (8), LOCOMAT (5)) providing symmetrical and repeatable gait-like training on a treadmill. Unfortunately, as these devices have limited degrees of freedom the walking practice is also somewhat restricted. In addition, these devices are expensive. A viable alternative to the mechanical powered orthosis is functional electrical stimulation (FES), which has a long tradition as an orthotic and therapeutic aid in the rehabilitation of walking after paraparesis (1). Direct stimulation of motor neurones, artificial activation of spinal neural circuits and stimulation of dermatomes have been employed successfully to augment artificially the movement of the affected lower extremity, usually during the swing phase. Years of clinical practice have shown that a single-channel peroneal stimulation is adequate assistance for correcting the condition of foot-drop or provoking flexion of the hip, knee and ankle in people with incomplete SCI.

It was therefore a natural step to combine the method of treadmill BWS training with single-channel FES, aiming to relieve the therapist from the task of manually assisting movement of the paretic extremity. Hesse (7) and Field-Fote (6) have clinically tested the above approach successfully and have demonstrated that such a practice has even better results than classical treadmill with BWS walking training when the application of FES is carefully controlled. Their experience has shown that such a combination of treadmill training with BWS and FES can only be effective when an experienced therapist manually controls the timing and intensity of the stimulation and visually evaluates the quality of the swings performed and continuously provides verbal feedback to the walking subject.

These findings motivated us to develop a system of multiple sensors for walking assessment and provision of cognitive feedback (4) with the goal of making the training process more objective and thereby also repeatable. The system developed enabled acquisition and assessment of the quality of the swing phase of walking while training on the treadmill. The system was tested in a person with an incomplete SCI level C6 injury and the results have shown that: (i) the subject was able voluntarily to improve the swing and thereby also improve their walking; and (ii) the timing of FES triggering was crucial, meaning that auditory feedback was also an important cue to the therapist who controlled the triggering and the intensity of the single-channel FES system augmenting the swinging of the paretic extremity.

The above work indicates the need for a fully automated sensory driven FES augmentation system that, based on the processed information acquired by the sensory system, apart from assessing and evaluating the swing phase and providing cognitive feedback signal to the patient, would also determine the proper timing and intensity of stimulation. The neuro-rehabilitation training incorporating automated FES augmentation, which depends on patients' performance, could further maximize the efficacy and quality of training.

METHODS

Control strategy

The backbone of the approach we propose consists of 2 feedback control loops, the cognitive feedback loop and the FES control loop (Fig. 1). In the first and more important feedback loop the patient is part of the strategy. The processed sensory information, which represents the estimated swing quality information, is provided as auditory feedback. The swing quality estimation is based on multisensor integration. Data assessed from knee goniometer, accelerometers and gyroscope are used to determine the numerical value (4), which represents a comparison with the desired reference swing. The reference swing movement could be captured either from the less affected lower extremity of the patient with SCI or determined as the trajectory of the lower extremity in a neurologically intact individual. The numerical value is provided to the patient as auditory feedback at 3 different levels. The levels are presented as 3 different frequencies. The low frequency indicates an adequate to poor swing, the middle frequency indicates a good swing and the high frequency represents an appropriately performed swing phase. **[Author - not clear which swing is best - a 'good' one or an 'appropriately performed' one?]** The feedback described is simple enough to be understood during walking and enables the patient voluntarily to improve the swing of his affected lower extremity. A more detailed description of the sensory system developed and processing algorithms utilized is provided in (4).

The FES control loop is based on the assessed sensory information and the swing phase quality. The sensors provide the information necessary to determine the moment of triggering of the electrical stimulation. Since we use single channel peroneal nerve stimulation to provoke the flexion response, i.e. simultaneous hip and knee flexion and ankle dorsiflexion, we have to make sure that the moment of triggering will take place before the initial swing phase. The swing phase can be divided into pre-swing, initial swing phase, mid-swing and terminal swing (9). We defined the appropriate moment of FES triggering using the information from the knee goniometer. When knee flexion occurs, the heel-off phase

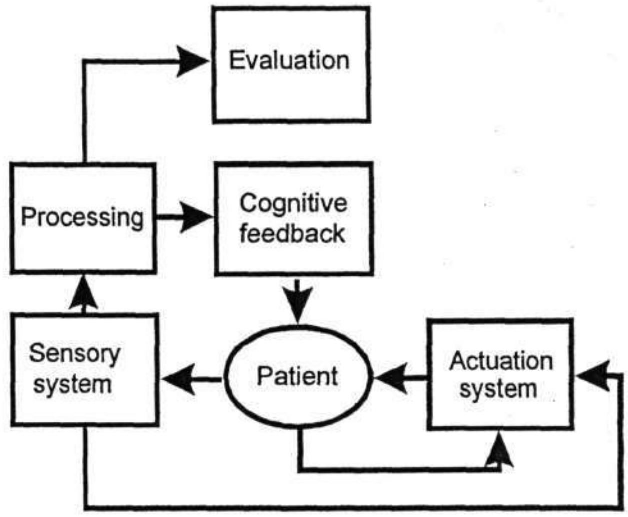

Fig. 1. The concept of FES gait re-education. Two loops are presented. The automatic actuation control loop is supported by a sensory system and the cognitive feedback loop includes the patient and his voluntary activities.

takes place and from numerous trials this appeared to be the most adequate triggering moment for peroneal stimulation.

Swing quality (4) is the term used to describe the correlation between the actual and the desired trajectory of the swinging limb and is the most important parameter used to control the stimulation amplitude. The stimulation amplitude is pre-set by the physiotherapist at the beginning of the session and depends on the observed deficiencies of a particular subject. During treadmill walking the quality of the swing phase is estimated and on the basis of this information the stimulation amplitude is adjusted. After every swing phase the quality estimate is stored. The user can pre-set the number of swings required for stimulation amplitude adjustment. When the pre-set number of good swings is performed, the stimulation amplitude is decreased. A succession of good swings means that the patient has managed to walk adequately; therefore the level of FES motor augmentation can be decreased. Conversely, in the case of a number of successive poor swings, the level of FES motor augmentation was not sufficient and did not allow the subject to perform adequately. Therefore, the stimulation amplitude is increased.

Instrumentation

The developed system (Fig. 2) was applied to a patient with incomplete SCI and tested in the rehabilitation centre. Most of the processing was managed by personal computer (PC, Pentium III 500 MHz), based on a Windows platform. The supervisor software, enabling set-up of the triggering and initial stimulation intensity, was programmed in Matlab/Simulink and C++. Signals were assessed by a multi-sensor device (4), consisting of 2 pairs of single cross-axial accelerometers and a gyroscope encased in a plastic housing. The rectangular device box was placed on the shank of the patient using 2 Velcro straps in the longitudinal axis aligned with the shank. Misalignment of the device in the longitudinal axis had no significant effect on results. Goniometers were placed at the ankle and knee joints in order to assess knee and ankle angle trajectories. The ankle goniometer had no other function than measurement, while the knee goniogram was used to trigger the FES and was also implemented in the sensory integration algorithm. The algorithm estimated the quality of the swing phase (4). The term "quality" here expresses the level of agreement with the chosen reference swing. The competent information is the acceleration trajectory of the foot. The user-friendly software provided several options to pre-set the required knee angle and thus the moment of triggering, the required knee flexion for good swing phase, the number of swings required for automatic stimulation amplitude adjustment and the parameters of the FES. We chose the clinically well-accepted single

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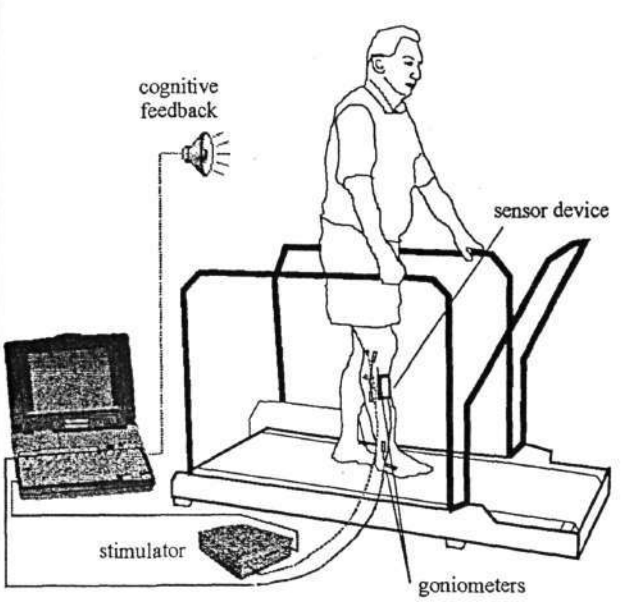

Fig. 2. FES re-education set-up. The multisensory system is attached to the shank of the patient. Data processing carried out by a personal computer, which provides cognitive feedback and controls the electrical stimulation, triggering and intensity.

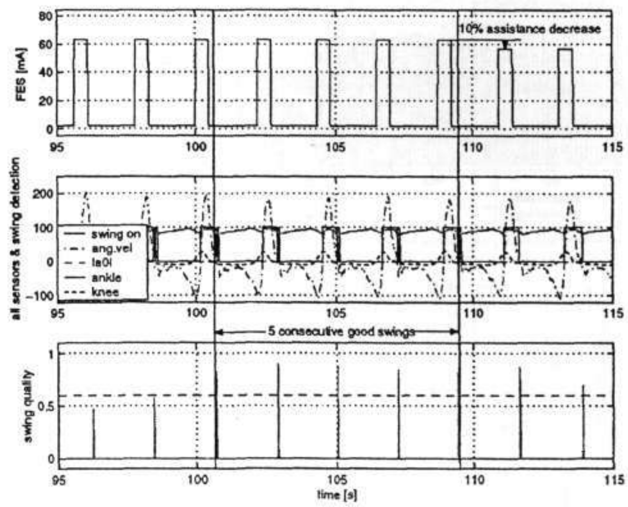

Fig. 3. The upper diagram shows the presence and the intensity (in mA) of electrical stimulation. The middle diagram shows all the assessed signals. "Swing on" presents the swing phase detection, which is based on gyroscope signal ("ang. vel") that assesses the angular velocity of the shank. The diagram shows both goniograms, the "knee" and the "ankle", and the acceleration ("|a0|") signal. In the lower diagram the swing quality is presented. After pre-set number (5) of consecutive good swings the stimulation intensity was decreased.

channel surface peroneal nerve stimulation with the following parameters, stimulation frequency 20 Hz, pulse width 200 μs and current 35 mA.

RESULTS

One male patient with incomplete SCI (height 1.75 metres, weight 84 kg, age 30 years) with C4–5 lesion, Asia classification C (4 months before measurement) was involved in the development, testing and case study validation of our approach. He was not able to walk due to inability to perform a swing movement without FES and was a regular peroneal stimulator, MicroFES^{TM} (The Jozef Stefan Institute), user for 1 month. MicroFES^{TM} is a 1-channel electrical stimulator, triggered by the shoe insole, with a pair of electrodes applied over the common peroneal nerve.

His FES assisted treadmill walking was considered satisfactory and was recorded as a reference gait pattern. This was done in the first session of the treadmill when the patient was walking with MicroFES^{TM} on both lower extremities. The goal of further treadmill training sessions, which were all performed on the same day, was to explore functioning of the completely automated sensory driven FES augmentation system. The computer controlled electrical stimulator, as described in (4), replaced the MicroFES on the patients' right lower extremity. The patient had to walk with the self-selected pre-set speed of the treadmill (0.5 km/h) and concentrate on the cognitive feedback signal. The goal of the subject, being the essential element of the first closed loop as shown in Fig. 1, was to perform in a way that resulted in good swings. The task of the second, automated closed loop, was to process data assessed by the multisensor device, and control the triggering moment and intensity of FES. At the beginning of the training session the physiotherapist pre-set the stimulation parameters, the required knee flexion in the swing phase and the required number (5) of good/poor swings for the automatic change of the stimulation intensity (10%). The intensity was also limited to prevent too much FES assistance.

The results show the assessed gait parameters, the presence of FES its intensity and the estimated swing quality during the selected training session (Fig. 3). The swing quality estimation took place at the end of each swing phase. The algorithm (4) was based on correlation with the reference signal (0 = poor, 1 = excellent) and returned a numerical value presented in the lower chart of Fig. 3. The figure also shows how the pre-set number of good swings (5) decreased the intensity of electrical stimuli (the upper diagram) by 10%.

The patient focused on the cognitive feedback audio signal and aattempted to improve his swing in order to maintain symmetrical walking. When the pre-set number of consecutive good swings had been performed, the stimulation intensity was decreased (Fig. 4). In contrast the stimulation level increased in the case of the pre-set number of poor swings. In all other cases (various successions of good, sufficient or poor swings) the stimulation intensity was not changed. The pre-set demand for swing classification was the swing quality value; during these measurements the swing quality value was set to 0.6. Figure 4 shows the gait re-education system performance and the FES assistance needed during walking. During the first phase of this gait training session the swing quality varied in such a way that the stimulation level remained unchanged. In the further course

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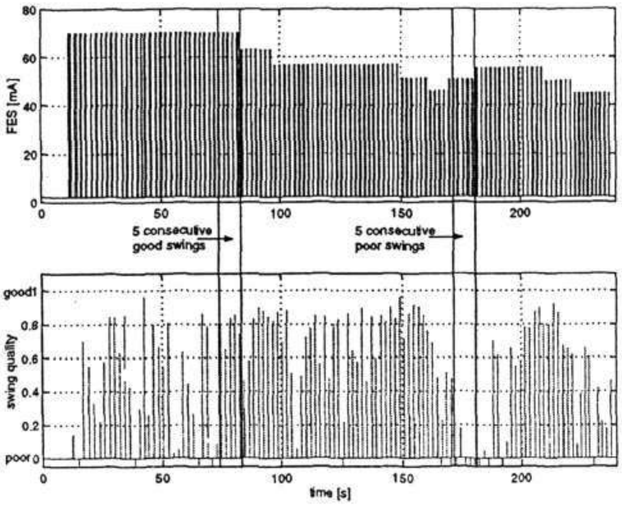

Fig. 4. The effect of the good/poor swings on the stimulation intensity. When the patients' swing phase became good/poor the requirement for FES assistance was decreased/increased.

of the session an improvement in performance can be observed resulting in a decreased level of FES support. In the interval from 150 and 200 seconds there is a decrease in swing quality which resulted in an increase in FES support, which enabled the subject to restore adequate swing performance and again reduce the level of FES assistance.

DISCUSSION

In our earlier development of a gait re-education system for use in incomplete paraplegia the aim was to involve the patient in the rehabilitation process (1) by using cognitive feedback supported by an artificial multisensor system. Triggering of the single-channel FES system augmenting the swing phase was left either to the patient or the therapist. We also tested the possibility of the walking subject adjusting the intensity of the electrical stimuli by a special control lever (3). Our earlier work (2) and the experiences of Hesse at al. (7) and Field-Fote (6) exposed the need for automated triggering and intensity adjustment of the FES system augmenting the swing in order to improve the repeatability of the training conditions where the level of FES support is adjusted to a level suitable for the walking subject's performance. The main focus of the work presented in this paper was therefore to develop the sensory driven automated FES augmentation system and explore the functioning of such a system in a clinical setting involving a neurologically impaired subject.

The results of our case study clearly show that the proposed approach of sensory driven automated FES augmentation of the swinging extremity can be successfully incorporated into gait treadmill training. The recordings presented in Fig. 4 clearly demonstrate the interaction of the walking subject and the state of the sensory driven FES augmentation system. From the gait performance it appears that the walking subject was indeed an active participant in the gait re-education process as the level of FES augmentation was reduced in the first phase of the training session and thereafter varied around the level that appeared to be needed. The main achievement of such training is that the walking subject receives not only the cognitive feedback information, which is needed to improve performance on a conscious level, but also relevant sensory feedback that is input to the spinal neural circuits at the spinal level as the level of FES support is related to the gait performance. The patient expressed that he was very contented with the device as he was released from the need to monitor visually the movement of his extremities. Thus he could move in a straight line and focus on the auditory feedback. At the same time the automatic control of the electrical stimulation provided the assistance required.

In conclusion, rather comprehensive and repeatable gait-training conditions are achieved which should have impact on the increased quality of walking **[Author - added 'of walking' - is this what you mean?]** and might also have shorten the duration of the gait re-education treatment required. By developing and testing the described gait re-education system we are in a position to evaluate further the presumed effectiveness of the approach in controlled clinical trials. Further development of the gait re-education system will focus on substitution of the knee goniometer information, needed to determine the triggering moment of the FES system. The triggering will be accomplished through adequate processing algorithm, which will derive the necessary information from the multisensor system. This will further simplify clinical utilization of the developed system.

ACKNOWLEDGEMENTS

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Izjavljam, da sem avtor te disertacije, ki je nastala kot plod raziskovalnega dela pod mentorstvom prof.dr.Tadej Bajda in somentorstvom doc.dr.Zlatka Matjačića. Vsa pomoč drugih sodelavcev je izkazana v Zahvali. Že objavljeni dosežki drugih avtorjev so navedeni v Literaturi.

Imre Cikajlo

Ljubljana, 19.maj.2003