筋骨格モデルを用いたFES歩行制御法の開発のための基礎的検討

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Fundamental Study on Development of Controlling Method of FES Gait by Using Musculoskeltal Model

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1. はじめに

我々の研究グループでは，機能的電気刺激（FES）による歩行再建を目指して，Cycle-to-Cycle 制御に基づく片麻痺者の遊脚期の制御に着目してモデルシミュレーションによる検討を行ってきた[1]．これまでに，片麻痺者の麻痺側の遊脚期の股関節角度と膝関節角度と足関節角度を制御するファジィ制御器を提案し，その有効性を示してきたが，麻痺側が遊脚期から立脚期へ切り替わる過程や麻痺側の立脚期の制御は検討していない．本研究では，麻痺側の立脚期も含めた FES による歩行再建を計算機シミュレーションで検討するため，片麻痺者を対象とし，麻痺側が遊脚期と立脚期の場合の各筋骨格モデルと床面モデルを構築した．これを用いて，順動力学問題を解くことで接地の状態まで含めた FES 歩行の計算機シミュレーションについて検討を行い，Cycle-to-Cycle 制御に基づく麻痺側の遊脚期の制御に改良を加えた．

2. 歩行モデルの構築

2－1．麻痺側が遊脚の場合の筋骨格モデル

麻痺側が遊脚の場合の骨格モデルは，図1のように，遊脚（麻痺側）の足，下腿，大腿，立脚（健側）の大腿，下腿の5つのセグメントと，それらを連結する左右の足関節と膝関節，股関節から構成した．なお，全ての関節は矢状面内のみに可動域を持つ蝶番関節とした．また，対象を歩行中の下肢に限定し，左右対称であること，立脚（健側）は動作を通して地面に固定であることを仮定した．頭部や体幹，上肢は1つの質点で表現し，その全質量を股関節上に集中させ，上体の向きは鉛直方向とした．各セグメントの質量は，セグメントの中央に1つの質点として集中させた[2]．遊脚の足部については，足の甲の部分に質点を集中させたセグメントとしたが，踵の位置は図2のように足関節からの長さを与えて決定した[3]．立脚の踵の位置も同様に決定した．関節トルクの符号は，全て反時計回りを正とした．セグメントの質量や長さなどのパラメータ値は文献[4]を用いた．

関節トルク（τ）は，電気刺激によるトルク（τ_{CE}）と受動粘弾性要素によるトルク（τ_P）の和として次

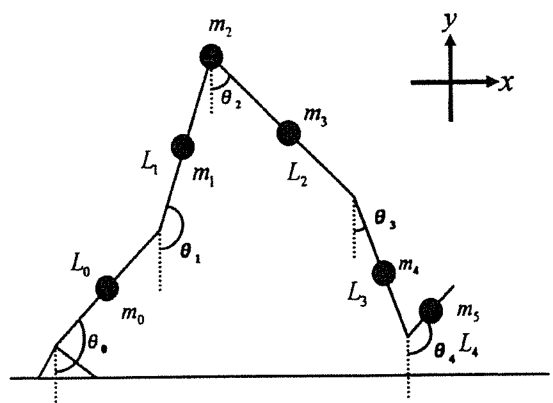


図1　遊脚期の骨格モデル（θ_0～θ_4：鉛直方向からの振れ角，L_0～L_4：各セグメント長，m_0～m_4：各セグメント質量を集中させた質点）

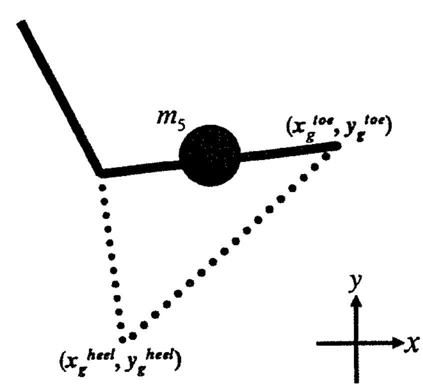


図2　遊脚（麻痺側）の足部における踵の位置および床面モデルの作用時の座標

式で求めた[5].

$$\tau = \tau_{CE} + \tau_P \tag{1}$$

τ_{CE}は，電気刺激により発生する筋収縮力F_{CE}とモーメントアーム（固定）との積により求められ，F_{CE}は，筋の活動度，長さ - 張力関係[6]，収縮速度 - 張力関係[7]，最大筋張力F_{max}で表されるHill型モデルにより求めた．なお，電気刺激による筋の活動度は，刺激強度に対する筋のリクルートメント特性[8]と活性化ダイナミクス[9]により求めた．受動粘弾性要素によるトルク（τ_P）は，文献[10]を参考にして各関節での運動毎に（2）式で記述し，これにより関節可動域も表現した．

$$\begin{aligned}\tau_p = &-k_{i1}\exp\{k_{i2}(\theta_i + k_{i3})\}\\ &+k_{i4}\exp\{-k_{i5}(k_{i6}+\theta_i)\} - c_i\dot{\theta}_i\\ &-k_{i7}\exp\{k_{i8}(\theta_i + k_{i9})\}\end{aligned} \tag{2}$$

ここで，k_{i1}～k_{i6}およびc_iは，関節毎に異なる係数であり，文献[11]を参考にした．k_{i7}～k_{i9}は，膝関節にのみ与える係数とし，それ以外の関節においては0とした．これは，麻痺側が遊脚の場合の骨格モデルが，麻痺側が立脚の場合の骨格モデルに切り替わった状態のシミュレーションを予備試験として行った結果，上体の質量による膝関節の過屈曲が生じたため，これまでの受動粘弾性要素の記述に補正を加えたからである．

運動方程式は**Lagrange**法により導出した．モデルに使用した筋は，各関節での各運動における主動筋となるものを選択した（表1）．

2－2．床面モデル

床面モデルは，作用点に働く力のx成分f_g^xとy成分f_g^yをつま先と踵において（3）式で表現した[11].

$$f_g^x = \begin{cases} -k^G(x_g - x_g^0) - c^G\dot{x}_g & (y_g \le 0)\\ 0 & (y_g > 0)\end{cases}$$

$$f_g^y = \begin{cases} -k^G y_g + c^G f_{\max}(-\dot{y}_g) & (y_g \le 0)\\ 0 & (y_g > 0)\end{cases} \tag{3}$$

$$f_{\max}(x) = \max(x,0)$$

ここで，k^G，c^Gは係数で踵とつま先で値が異なるとし，x_g, y_gは踵とつま先においてそれぞれx_g^{heel}, y_g^{heel}, x_g^{toe}, y_g^{toe}として与えられる床反力作用点位置(図2)，x_g^0は接地した瞬間の床反力作用点位置である．ここで，立脚の踵のy座標を0として与え，遊脚（麻痺側）の足部における床反力作用点位置のy座標が0以下ならば，床面モデルが作用するものとした．この床面モデルによって得られた床反力を（4）式により等価関節トルクに変換した[12].

表1　モデルに含まれる筋とその作用

主動筋	作用
ヒラメ筋 (Sol)	足関節の底屈
前脛骨筋 (TA)	足関節の背屈
腓腹筋 (Gas)	足関節の底屈，膝関節の屈曲
大腿直筋 (RF)	股関節の屈曲，膝関節の伸展
広筋群 (Vas)	膝関節の伸展
ハムストリングス(大腿二頭筋短頭・大腿二頭筋長頭) (Ham)	股関節の伸展，膝関節の屈曲
腸腰筋 (IL)	股関節の屈曲
大臀筋 (GM)	股関節の伸展

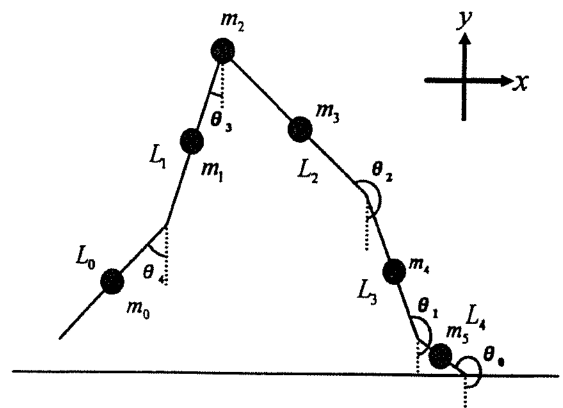


図3　立脚期の骨格モデル（θ_0～θ_4：鉛直方向からの振れ角，L_0～L_4：各セグメント長，m_0～m_4：各セグメント質量を集中させた質点）

$$\tau = J^T F \tag{4}$$

ここでτは各関節トルクのベクトル，Jはヤコビ行列，Fは作用点に働く力のベクトルである．

2－3．麻痺側が立脚の場合の筋骨格モデル

麻痺側が立脚の場合のモデルは，基本構造は麻痺側が遊脚の場合の骨格モデルと同じであるが，遊脚（健側）の足部は省略した（図3）．なお，麻痺側の立脚期においては足のつま先位置は動作を通して地面に固定であることを仮定した[2]．セグメントの質量や長さなどのパラメータ値は麻痺側が遊脚期の場合と一致させた．

立脚（麻痺側）の足部においては，つま先を固定としたため，床面モデルを立脚（麻痺側）の踵のみに付けた．ただし，踵はつま先を中心にした円軌道上のみを運動することになるため，床反力としては(3)式のf_g^yのみが作用することとし，踵の床反力は，その作用点のy座標がつま先のy座標以下で作用するものとした．一方，遊脚（健側）においては足部を省略しているため，床面モデルを遊脚（健側）の踵とつま先に付けることができない．そこで，遊脚（健側）の足底全体で受ける床反力として計算され

るものとし，遊脚（健側）の下腿の先端に床面モデルを付け，遊脚（健側）の下腿の先端の y 座標が立脚（麻痺側）の足関節の y 座標以下になると床面モデルが作用するものとした．

3．計算機シミュレーション

3－1．方法

前述の筋骨格モデルを利用し，ファジィ制御を用いた Cycle-to-Cycle 制御による麻痺側の遊脚期の制御[1]に関する計算機シミュレーションを行った．片麻痺者を想定しているため，立脚（健側）の各関節角度は健常者の歩行における立脚の股関節位置と膝関節位置の軌道を再現する角度をモデルに合わせて修正したものを入力し，遊脚の各関節角度の初期値は健常者の歩行における対応する関節角度と一致させた．

Cycle-to-Cycle 制御では，電気刺激パルス列を印加する時間 TB[s]を調整し，遊脚期における重要な点での角度を制御する．つまり，関節の最大屈曲角度などを制御対象とし，現在の cycle におけるその角度と対応する目標角度の誤差をフィードバック信号として次の cycle に送り TB[s]を調整し，電気刺激が印加される．本研究で制御する関節角度は，歩行の遊脚期における股関節の最大屈曲角度，膝関節の最大屈曲角度，足関節の最大背屈角度と最大底屈角度，遊脚期の最終姿勢（着床）における各関節の角度とした[1]．電気刺激スケジュールは，先行研究によって歩行制御が適切であると評価されたもの（図4）を用いた[13]．したがってファジィ制御を用いた Cycle-to-Cycle 制御による麻痺側の遊脚期の制御が開始すると，足関節の最大底屈角度を目標とする Gas への電気刺激，膝関節の最大屈曲角度を目標とする Ham への電気刺激，股関節の最大屈曲角度を目標とする Il への電気刺激が印加される．次に，足関節の最大底屈角度が検出されると，最大背屈角度を目標とする TA への電気刺激が印加される．そして，足関節の最大背屈角度が検出されると，最終姿勢に向けて TA と Sol への電気刺激が印加され，膝関節の最大屈曲角度が検出されると最終姿勢に向けて RF と Vas への電気刺激が印加され，股関節の最大屈曲角度が検出されると最終姿勢に向けて Il への電気刺激が印加される．

ここで，本報告で構築した筋骨格モデルでは，先行研究における遊脚の受動粘弾性要素のパラメータ値[9]では脱力状態において不適切な角度となることが予備的なシミュレーション実験で確認されたため，遊脚（麻痺側）の足関節の受動粘弾性要素のパラメータ値を変更した[11]．その結果，Cycle-to-Cycle 制御による足関節制御において最大底屈角度が検出されず，つま先が初期姿勢から最終姿勢に向けて重力の影響を受けて単調に底屈した．これに対して，健常者の歩行では，膝関節の遊脚期の最大屈曲角度の約 70%に膝関節角度が達すると，足関節の最大底屈が生じていたことから，これを参考にして足関節制御器を改良した．つまり，*n* 歩（*n* cycle）目で足関節の最大底屈角度が検出されなかった場合，次の *n* +1 歩（*n* +1 cycle）目では，初期姿勢から膝関節の最大屈曲角度が検出されるまでの間で，膝関節角度（$\theta_{knee}[n+1]$）が一つ前の cycle における最大膝屈曲角度（$\theta_{knee\_max}[n]$）に対する 70%の角度に達すると，足関節の最大背屈角度に向けて TA への電気刺激が印加されるように制御を変更した．ただし，$\theta_{knee}[n+1]$が$\theta_{knee_max}[n]$の 70%に達する前に足関節の最大底屈が検出された場合には，それを優先させた．

最初に，これまでの研究[1]と同様に，麻痺側の遊脚期のみの制御について，各筋の TB[s]の初期値を 0[s]として 50 歩分のシミュレーションを行った．被刺激筋は先行研究[1]との比較のため，表１における GM 以外とし，電気刺激を印加した．

次に，麻痺側の遊脚の終了において，麻痺側の足部が床面に接する状態の動作を確認するため，床面モデルを含めてファジィ制御を用いた Cycle-to-Cycle 制御に基づく麻痺側の遊脚期の制御に関する計算機シミュレーションを行った．

最後に，麻痺側が遊脚期から立脚期に切り替わる過程に関する計算機シミュレーションを試みた．遊脚（麻痺側）の足部が床面とほぼ平行になった時点で遊脚（麻痺側）の足部のつま先が固定されて，麻

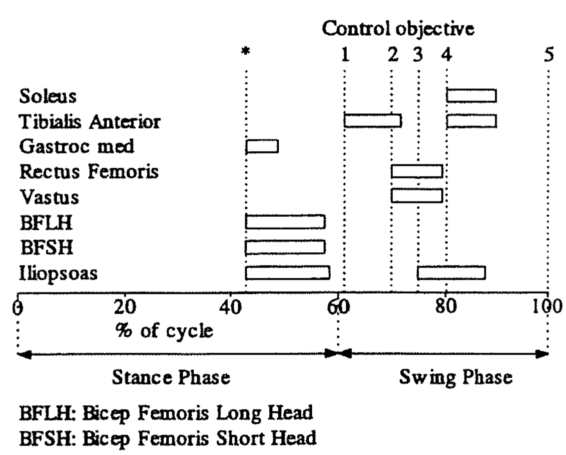


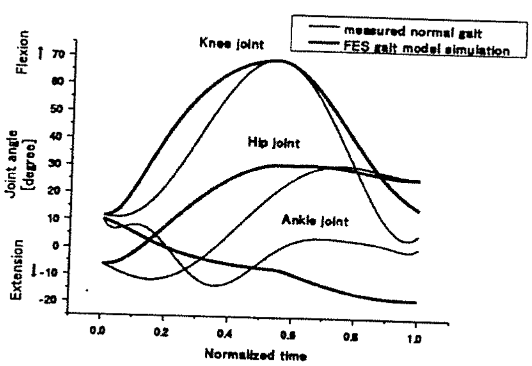
図４　先行研究における電気刺激スケジュール[13]
（*：遊脚開始，1：足関節の最大底屈角度の検出時，2：膝関節の最大屈曲角度の検出時，3：股関節の最大屈曲角度の検出時，4：足関節の最大背屈角度の検出時，5：遊脚の最終姿勢（着床））

痺側が立脚期に切り替わることとし，その時点での立脚（健側）と遊脚（麻痺側）の各関節の角度，角速度を麻痺側が立脚の場合のモデルにおける，遊脚（健側）と立脚（麻痺側）のそれぞれの初期値として与えた．なお，麻痺側（立脚）には電気刺激を与えず，また健側（遊脚）も床面モデルから力を受けるものの能動的な力を出さないようにして計算機シミュレーションを行った．

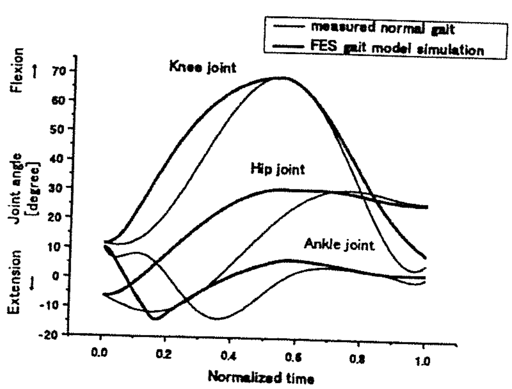
なお，計算機シミュレーションの時間ステップは 5×10^{-6}[s]とし，制御結果は20[ms]ごとに測定した．微分方程式の解法として4次のルンゲクッタ法を用いた．

3－2．結果

最初に，麻痺側の遊脚期のみのFES制御による



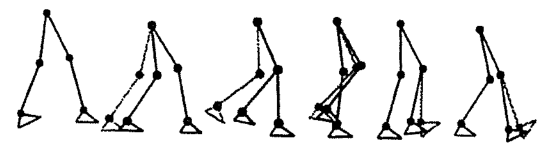
(a)従来の方法



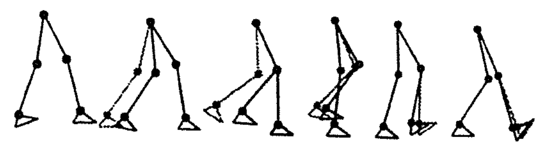
(b)改良型の方法

図5　各関節の角度軌跡．股関節角度：鉛直方向から屈曲方向を正，膝関節角度：大腿部の延長線と下腿部の為す角について屈曲方向を正，足関節角度：足部の甲と下腿部の延長線が為す角について90°を0°として屈曲方向を正としてプロットした．

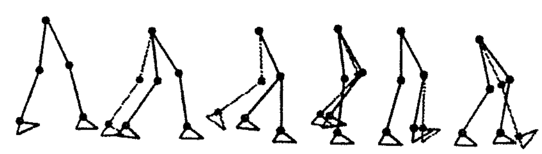
結果について，図5に各関節角度の軌跡，図6にstick picture，図7に得られた電気刺激バーストパターンを示す．図5(a)，図6(a)，図7(a)は従来の方



(a)　従来の方法，床面モデル無し

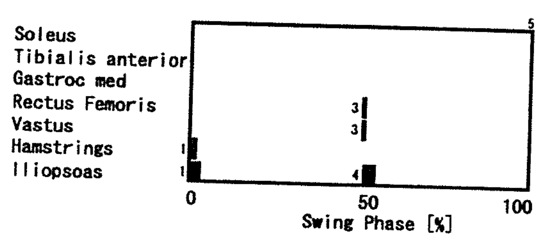


(b)　改良型の方法，床面モデル無し

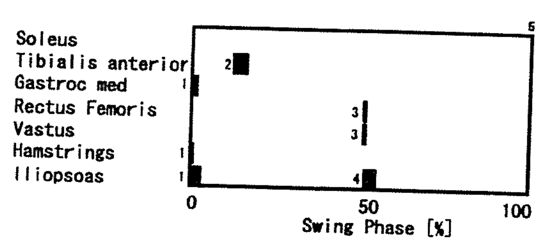


(c)　改良型の方法，床面モデル有り

図6　遊脚の制御の様子(遊脚期の200 ms毎のstick picture，黒：モデル，灰：健常者)



(a)　従来の方法



(b)改良型の方法

図7　遊脚期を適切に再建する電気刺激バーストパターン（1：遊脚開始，2：足関節最大底屈，3：膝関節最大屈曲，4：股関節最大屈曲，5：遊脚終了)

図 8　麻痺側が遊脚期の場合のモデルから麻痺側が立脚期の場合のモデルへと切り替わった後のシミュレーション結果（モデルの切り替わり後 65ms 毎の stick picture．黒：麻痺側，灰：健側）

法[1]による制御結果であり，筋骨系の特性によっては，足関節の最大底屈角度が検出されないことが確認された．この問題点を改善するため，足関節の制御器を前述のように改良した場合の制御結果が図 5(b)，図 6(b)，図 7(b)であり，麻痺側の足関節の制御が適切に行われていることが分かる．しかし，本報告でのモデルによる計算機シミュレーションでは，図 7(b)において足関節の最大背屈位から踵の着床に向けて，TA と Sol への電気刺激は印加されなかった．先行研究での結果では，TA と Sol への電気刺激の印加は必要であったが，本モデルを用いた結果では，着床時の足関節の角度誤差が非常に小さくなったため不要になったと思われる．

次に，図 7(b)の刺激バーストパターンによる制御の最終姿勢において，着床する状態についてのシミュレーション結果を図 6(c)に示す．図 6(b)では，床面モデルが無いために，麻痺側の遊脚期の終了時点での運動の状態により各関節角度が変化する．そのため，股関節角度は大きく変わらず，膝関節が伸展位を維持している．図 6(c)では，踵の着床により，床面モデルが作用し，遊脚（麻痺側）の股関節と膝関節が屈曲して，さらに床反力により足関節の底屈が生じ，足底が床面に対して平行になっている．

最後に，足底が床面に接した後に麻痺側を立脚期とした場合のモデルへ切り替えた結果を図 8 に示す．モデル切り替え後は，立脚（麻痺側）には電気刺激を与えず，また遊脚（健側）も床面モデルから力を受けるものの能動的な力を出さないようにしたため，モデル切り替え後に両側の膝関節が屈曲しており，妥当な結果が得られている．

4．考察

ファジィ制御を用いた Cycle-to-Cycle 制御による結果で，本報告で構築した筋骨格モデルの特性が変わった場合，先行研究の制御器では足関節の最大底屈角度が検出されない場合があることが新たに確認された．このような足関節の応答は，比較的ゆっくりとした動作を行う麻痺患者の歩行において実際に起こりえると考えられる．そこで本報告では，足関節の最大底屈角度が検出されない場合も考慮して足関節の制御器を改良し，おおむね妥当な歩行が再建されることを確認した．先行研究の制御器では，全ての関節において最大屈曲角度や最大背屈角度などが検出されることが前提となっており，それらが検出されなかった場合の関節角度の制御は考慮されていない．そのため，他の関節の制御器に関しても，最大屈曲角度や最大背屈角度などが検出されない場合を考慮して改良する必要があると考えられる．

遊脚（麻痺側）の足関節角度が単調に底屈する様子，床反力の作用の様子，麻痺側が遊脚の場合のモデルから麻痺側が立脚の場合のモデルへと切り替わった後の転倒の様子から，構築されたモデルの動作は妥当であると考えられる．しかし，より実際に近い状態を表現するためには以下の点が課題であると考えられる．まず，床反力を等価関節トルクに変換する場合，麻痺側が遊脚から立脚へ切り替わる過程では，つま先が固定されると，麻痺側が遊脚の場合のモデルで働いていたつま先の床面モデルの作用が終了するので，切り替え後の初期値には，各関節の等価関節トルクを与えていないことである．また，一般的には麻痺側が立脚期の場合は松葉杖などの支えによる力が上体を介して伝えられるが，本報告における骨格モデルでは上体の姿勢が鉛直方向であると仮定しており，実際の姿勢と異なっている．さらに，床面モデルの出力値を評価していないため，実験条件との対応が不十分である．上記のような点が動作に影響を与えることが考えられる．

5．まとめ

本報告では，歩行の FES 制御法の開発に利用することを目的とした筋骨格モデルと床面モデルの構築を行い，これまでの麻痺側の遊脚期のみの制御から立脚期も含めた制御のシミュレーションに拡張するための基礎的検討を行った．床反力の作用や，麻痺側が遊脚の場合のモデルから麻痺側が立脚の場合のモデルへ切り替わる過程の結果から，本モデルがおおむね妥当であり，本モデルを用いて一定の歩行状態を表現できると思われる．また，麻痺側の遊脚期の制御では，足関節の制御において最大底屈角度が検出されない場合の対策を加えて改良し，麻痺側の足関節の制御が適切に行われることを確認した．今後は，麻痺側が立脚期の場合の FES 制御法の開発を進めるとともに，より実際の状態に近いモデルを実現するための検討を行う予定である．

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表面電気刺激による皮膚感覚を用いたパターン提示に関する検討
―電気刺激感覚の残存についての基礎実験―

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Study on pattern presentation using cutaneous sensation elicited by surface electrode stimulation: Basic experiment of remaining electrocutaneous sensation

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1．はじめに

皮膚感覚は視覚や聴覚に代わる情報伝達の手段として重要であり，例えば，義肢や機能的電気刺激（FES）を使用する場合には，制御の命令や状態を使用者に示す際に有効になる．これには，機械的な振動を用いる方法と電気刺激を用いる方法がある．電気刺激による手法では，提示装置として電極を装着するだけで良いため，機械的手法に比べ小型で簡単な提示装置にすることができる．そのため，本研究では，電気刺激により生じる感覚を用いて情報提示を行う手法に着目する．

梶本ら[1]は，指先に装着した電極で電気刺激感覚を提示する触覚ディスプレイを提案している．この手法では，提示装置が指先を覆ってしまうため，手を使う作業での情報提示への応用は困難である．また，触覚を提示するのに最も適した部位は指先であると言われているが[2]，義肢使用者や麻痺患者に関しては，残存する手指などは他の用途や装置の操作等に使用するため，本研究で目的とするような情報を提示する部位としては適切とは言えない．一方，金ら[3]は，前腕に電極を配置して，電気刺激パターンをある単語に結びつけて情報の伝達を行う方式を提案している．この手法は，刺激パターンを単語と対応させるため，提示するパターン数を多くできないといった課題が挙げられる．

本研究では，単語や意味情報を直感的に与えることに着目し，装置操作や作業の妨げとならない前腕部を提示部位として，文字のようにそれ自体に意味のある情報を皮膚電気刺激感覚により提示する方法を検討する．Kaczmarek ら[4]は，マトリックス状の電極を指先でなぞることで静的なパターンを提示する方法を検討している．しかし，この手法では，4 種類だけの提示パターンであるにも関わらず，約 70～78％程度の正答率であり、実用的であるとはいえない．これに対し本研究では，図 1 に示すマトリックス状に配置した電極を用いて，電気刺激を印加する電極を変えることで生じる移動感覚により図 2 に示す 16 種類のパターンを提示し，それらの識別について検討した[5]．その結果，パターン認識における間違いにいくつかの共通する傾向が見られ，提示パターンを工夫することで，多数のパターンを提示した場合でも，Kaczmarek らの結果と同程度の認識精度が得られることまでは示唆された．

本報告では，提案する手法について，さらに認識精度を改善するため，誤認識の原因について検討した．特に，以前に得られたパターン認識における間違いの傾向の中で，2 本の線上を順に移動するパターン[9]と[10]を，それぞれパターン[14]や[16]，パターン[13]や[15]といった，L 字や V 字のパターンとして認識する傾向，及び提示時間の異なるパターン同士を間違えるといった傾向に着目した．そして，これらのパターン誤認識の原因の１つと考えられる，電気刺激感覚の残存の影響について実験により検討した．

2．単一電極での電気刺激による感覚の残存の影響

被験者として，前述の 2 つの間違いの傾向が多く見られた 2 名（健常被験者C，E）を選び，実験を行った．パターン認識実験に用いた図 1 の電極を使用し，導電性のペーストを電極に塗

付して，被験者の左前腕前面の中央に装着した．電気刺激を印加する電極は，図1の電極の中央1行の3つの電極とし，1秒間の電気刺激を，0s，0.1s，0.3s，0.5s，0.8sいずれかの刺激間隔で2回与えた（図3）．電気刺激は，パルス幅200μs，パルス周波数100Hzの単相性矩形波[6]とした．被験者には，2回の電気刺激の間の時間間隔の有無を回答させた．この実験を各電極に対して5つの刺激間間隔を各8回ずつ，計120回をランダムに提示した．

実験の結果，2名の被験者とも時間間隔のあった場合は100%判別することができた．そのため，一つの電極に対する刺激に関しては，電気刺激感覚の残存の影響はほとんど見られないと考えられる．

3．　移動感覚における感覚の残存の影響

3．1　方法

パターン[9]と[10]を用い，移動する刺激に関して，刺激感覚の残存の影響を調べた．被験者として前述の間違いの傾向のうちのひとつ，パターン[9]，[10]に関する間違いが多く見られた2名（健常被験者A，C）を選び，実験を行った．実験では，最初に各電極について被験者がパターンを判断しやすい刺激強度を設定した．被験者には，提示するパターンについて具体的な形や刺激のタイミングは説明せず，何らかのパターンを2回提示することを伝えた．そして，1番目と2番目の提示パターンが同じかどうか，また，それぞれどのような形のパターンとして感じられたかを回答させた．1番目のパターンとしては，パターン[9]（もしくは[10]）を提示し，2番目のパターンとしては，1番目に提示したパターンの1本目と2本目の刺激の間に，0s，0.1s，0.3s，0.5s，0.8sの刺激間間隔を含めたパターンをのいずれか提示した．

実験は，最初にパターン[9]について10回行った（5種類の刺激間間隔を各2回）．この時，刺激間間隔をランダムに選択し提示した．次に，同様の実験をパターン[10]について10回行い，再びパターン[10]，パターン[9]の順に測定を行った．なお，10回の測定が終了する毎に，休憩をとり，刺激強度の再調整を行った．このようにして，各パターンについて20回，計40回の測定を行った．

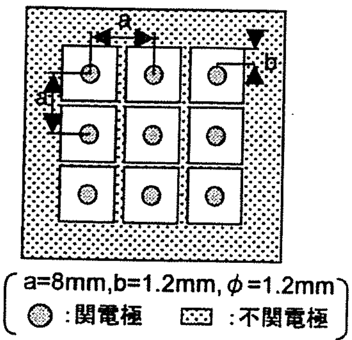

図1．パターン認識実験に用いた電極

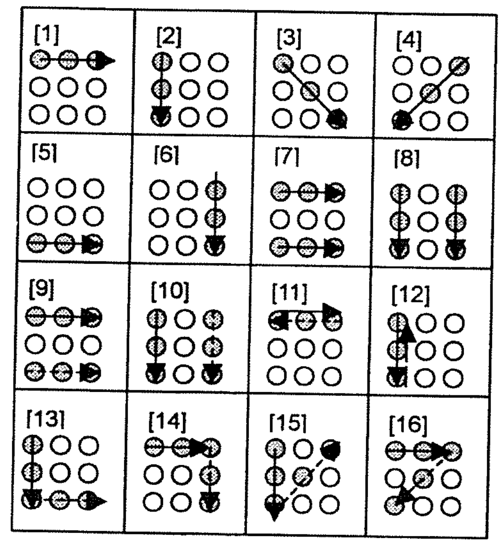

図2．パターン認識実験で用いた提示パターン

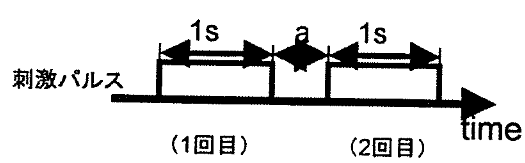

図3．2回の電気刺激の提示方法
（刺激間間隔 a=0.1，0.3，0.5，0.8s）

3．2　結果

被験者Aの場合，刺激間間隔によらず，2回の提示パターンは同じパターンであると回答したが，パターン[10]の場合のみ，1番目と2番目の両方とも，図2で示すL字のパターン

表1．1番目と2番目の電気刺激提示の違いに対する被験者の回答の割合（%）

被験者	パターン[9]			パターン[10]		
	刺激間間隔	パターン	両方	刺激間間隔	パターン	両方
A	100.0	0.0	0.0	100.0	0.0	0.0
C	100.0	0.0	0.0	41.2	52.9	5.9

表2．パターン[9], [10]に対する誤認識率 (%)

被験者	パターン[9]					パターン[10]				
	0s	0.1s	0.3s	0.5s	0.8s	0s	0.1s	0.3s	0.5s	0.8s
A	0.0	100.0	75.0	16.7	0.0	0.0	42.9	0.0	33.3	0.0
C	20.0	20.0	0.0	0.0	0.0	71.4	33.3	0.0	0.0	0.0

[13]として認識した回答があった（刺激間間隔0s：2回，0.8s：1回）．被験者Cでは，パターン[9]の場合には，刺激間間隔によらず，パターン[9]と回答したが，パターン[10]については，パターン[14]のような軌跡を持つパターンと回答したのが1回(0.5s)，左から右へ点刺激が2〜4点移動した後，右下へ1，2点の点刺激を感じる，というパターンを認識した回答が10回（0s：4回，0.1s：2回，0.3s：1回，0.5s：1回，0.8s：2回)，何かの形にはなっていなかったが点刺激を何点か感じ，それらの刺激に時間的なずれがあったように感じた，という回答が7回（0s：2回，0.1s：1回，0.3s：2回，0.5s：1回，0.8s：1回）あった．

表1に，被験者が1番目と2番目の刺激を「違う」と回答した場合について，刺激間間隔，パターンのうちどれが異なるために「違う」と回答したかを，各パターンの回答のうち「違う」と回答した数に対する割合としてまとめた．被験者Aは，1番目と2番目の提示パターンについて，パターン[9]，[10]いずれの場合も刺激間隔が異なるために違うと回答している．一方，被験者Cは，パターン[9]については刺激間間隔が異なるために違うと回答しているが，パターン[10]については，刺激間間隔，パターン，もしくはその両方を違う理由として回答している．

表2に，各刺激間間隔についての誤認識の割合を示す．被験者Aは，パターン[9]では，刺激間間隔0〜0.3sで，刺激間間隔の検出がかなり困難であり，0.5sでも検出ができない場合があった．また，パターン[10]についても，0.3sでは認識できているが，刺激間間隔0.5s程度までは刺激間間隔を検出しにくいといえる．一方，被験者Cは，パターン[9]については0s，0.1sで刺激間間隔の有無が曖昧になっているが，0.3sより大きい場合には，パターン[9]，[10]ともに提示した刺激の差異を正しく認識できているといえる．

3．3　考察

今回の実験では，最初のパターンを提示した後に，刺激間間隔だけが異なる同一のパターンを提示したため，被験者Cのパターン[10]の場合を除いて，刺激間間隔の違いにだけ着目して提示パターンの差異を認識したと考えてよいと思われる．実験後，「パターンより時間間隔の違いがはっきり分かるので，そちらばかりを意識してしまう」というコメントが両被験者から得られており，表1に示したように，被験者Cのパターン[10]の場合を除いて，刺激間間隔の違いに着目して提示された電気刺激の違いを識別しているので，被験者は，先入観なく刺激間間隔の有無を検出したと推測される．

被験者Cは，パターン[10]に対し，刺激間間隔が0sの場合も，異なるパターンであると回答する割合が多く，また，刺激間間隔によらず，パターン[14]に似たパターンや，いくつかの点刺激の集まりとして認識することが多かった．このことから，パターン[10]に用いる6つの電極のうちいくつかだけが認識され，判断していたと推測される．6つの電極すべてが同様に

知覚されない原因については，電気刺激感覚の残存により影響が生じること，神経や皮膚分節が，前腕に対し縦方向に伸びているため，縦方向に対する，電気刺激感覚が知覚しにくいことなどが考えられる．

表２より，被験者Ａは刺激間間隔が 0.5s の場合まで 0s の場合と混同する結果となったことから，実際の提示時間より 0.1～0.5s 長く電気刺激を知覚する場合があることが示唆される．よって，パターン提示において，１つの電極の刺激時間が 0.33s であることから，電極約２個分の刺激が正確に知覚されなくなる可能性があると推測される．したがって，パターン識別における間違いの原因として，以下のような二つのことが考えられる．

① パターン[9]や[10]について，２本の刺激の中間付近が曖昧になってしまうため，刺激の開始と終了に着目してＬ字のようなパターンとして判断したり，後半の刺激がそれ以前の刺激の影響を受けて曖昧になり，連続的なＶ字のようなパターンとして判断してしまう．

② 実際の提示時間より長く感じられてしまうため，提示時間の違うパターンとして判断してしまう．

被験者Ａがパターン[10]をパターン[13]と認識した原因は、上記①と同様であると考えられる．

４．結論

本報告では，皮膚電気刺激感覚を用いて，文字情報に相当するような意味のある感覚パターンを提示する方式を開発するため，16 種類の提示パターンの認識における間違いの原因について，電気刺激感覚の残存の影響に着目し，実験により検討を行った．実験結果から，電気刺激感覚の残存がパターンの認識に影響を与えている可能性が示唆された．この結果は今後，本手法によるパターン提示を行う上で，刺激パターンを改良するための知見として有用であると考えられる．

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Short Paper

A Test of Stimulation Schedules for the Cycle-to-Cycle Control of Multi-joint Movements in Swing Phase of FES-induced Hemiplegic Gait

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Abstract This paper described a computer simulation test of six different stimulation schedules for the cycle-to-cycle control of swing phase of functional electrical stimulation (FES) induced hemiplegic gait. The stimulation schedules were evaluated in controlling the hip, the knee and the ankle joint movements on the point of view of acceptable quality of the gait that was similar to the natural gait pattern. Five stimulation schedules were knowledge-based stimulation schedules and one stimulation schedule was an EMG-based stimulation schedule. Two acceptable stimulation schedules were found by the evaluation. Results of this study showed that combination of the timing pattern of muscle activation and knowledge of joint movements and muscle function is necessary in design of stimulation schedule for FES gait. Co-activation of the ilopsoas, the hamstrings and the vastus muscle at the beginning of swing phase and that of the tibialis anterior and the soleus at the end of swing phase were found to be effective in controlling swing phase. The knowledge-based generation of stimulation schedule would be effective and necessary in clinical application.

Key Words: FES, Cycle-to-cycle control, Hemiplegic gait, Stimulation schedule

1. Introduction

The cycle-to-cycle control is a control method of FES gait that regulates stimulation burst duration of stimulation pulses of a current cycle of gait based on the performance of previous cycles, whereas pulse width, amplitude and frequency are fixed. The target joint angles are some important points of the joint angle during a particular gait phase (e.g., maximum joint angles and joint angles at initial contact). In this method, parameter of the stimulation pulses that has direct relationship to the achievement of the target joint angle is the stimulation burst duration. Fixing of the amplitude, the pulse width and the frequency of the stimulation pulses is aimed to generate a stable muscle force. By regulating the stimulation burst duration, the controlled joint angle is affected to reach the target joint angle. Since the trajectory-based control method does not show effectiveness in generating smooth joint angle trajectories, the cycle-to-cycle control is expected to be an alternative.

The cycle-to-cycle control showed capability to realize target joint angle in single-joint control of gait induced by functional electrical stimulation (FES) [1,2]. Since implementation of the multi-joint control is essential for practical use of the cycle-to-cycle control, we have implemented the cycle-to-cycle control in fuzzy controllers and have found that it would be effective in controlling multi-joint movements during swing phase of FES gait [3]. A test of the quality of FES-induced gait would be required in the next stage for practical application of the cycle-to-cycle control.

In the cycle-to-cycle control, sequence of the muscle stimulation is arranged in a stimulation schedule, which is relevant to quality of the FES gait. In our previous studies, the stimulation schedule was created based on knowledge of joint movements during a particular gait phase and muscle functions. As the other method of creating stimulation schedule, the EMG pattern can be used. In this paper, in order to test the concept of design of the stimulation schedule, we tested six different stimulation schedules for the cycle-to-cycle control for multi-joint control of swing phase of FES-induced hemiplegic gait including an EMG-based stimulation schedule. The other five stimulation schedules that were knowledge-based stimulation schedules were included in order to test effectiveness of the knowledge used in the design of the previous stimulation schedule. The test was performed in computer simulation using an electrically stimulated musculo-skeletal model. Each stimulation schedule was evaluated by comparing the estimated joint angle trajectories of FES control to measured trajectories of the normal gait.

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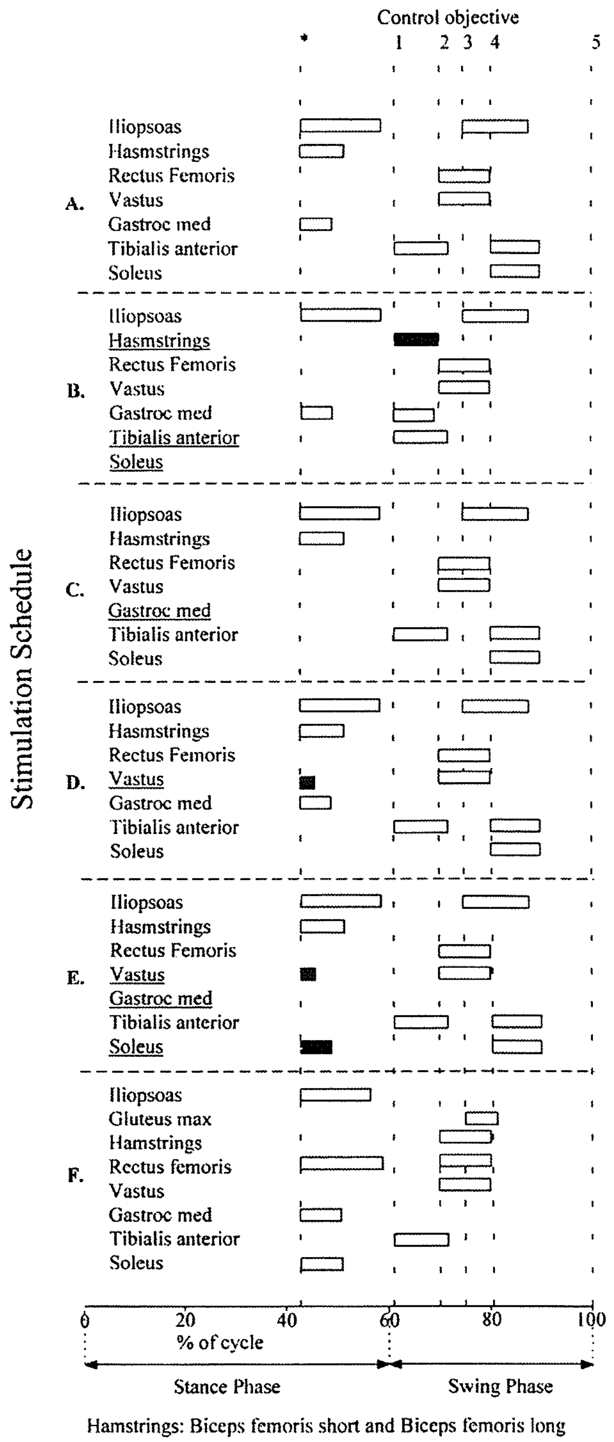

Fig.1 Six different stimulation schedules for multi-joint control. *: beginning of stimulation (maximum hip extension angle, maximum knee extension angle, and maximum ankle dorsiflexion angle at the end of stance phase). Control objective: 1: maximum ankle plantar flexion angle, 2: maximum knee flexion angle, 3: maximum hip flexion angle, 4: maximum ankle dorsiflexion angle, and 5: maximum knee extension angle and hip and ankle angles at initial contact. The underlined muscle name and filled stimulation timing in the stimulation schedules B-E are the points to be tested according to the stimulation schedule A. See text in detail.

2. Methods

2.1 Stimulation Schedule

Fig.1 shows the six stimulation schedules for the cycle-to-cycle control that were tested in this paper. Beginnings of the muscle stimulation were at the maximum hip extension, maximum knee extension or maximum ankle dorsiflexion angles at the end of stance phase. In normal gait, those maximum joint angles usually occur at different time in a cycle of gait. In order to facilitate the computer simulation, we assumed these maximum joint angles occurred simultaneously.

The stimulation schedule A was designed in the previous study [3]. Stimulation of the iliopsoas, the hasmtrings (the biceps femoris long head and the biceps femoris short head) and the quadriceps (the rectus femoris and the vastus muscles) was to produce the joint movements reaching the targets of maximum hip flexion angle, maximum knee flexion angle, and maximum knee extension angle, respectively. Stimulation of the gastrocnemius medialis and the tibialis anterior muscles was to produce ankle joint movements reaching the targets of maximum ankle plantar flexion angle and maximum ankle dorsiflexion angle, respectively. The stimulation of the hamstrings formed co-activation of the biceps femoris long head and the iliopsoas, which was aimed to avoid excessive hip flexion at the beginning of swing phase. In order to keep the hip joint in flexion and reaching the target of hip joint angle at initial contact, the iliopsoas was stimulated again after the hip joint reached the target maximum hip flexion angle. The co-activation of the tibialis anterior and the soleus was to prevent unstable movements of the ankle joint at the end of swing phase.

The following five stimulation schedules were designed in order to test design concept in the stimulation schedule A. In the stimulation schedule B, the hamstrings were stimulated after the ankle joint angle reached the target of maximum plantar flexion, in order to reduce excessive knee flexion caused by simultaneous stimulation of the hamstrings and the gastrocnemius medialis as seen in movement developed by the stimulation schedule A [3]. We also omitted the co-activation of the tibialis anterior and the soleus in order to test the significance of that. The stimulation C was aimed at testing of effect of omitting the stimulation of the gastrocnemius medialis at the beginning of control on knee flexion. Effect of the co-activation of the vastus muscles with the hamstrings in the knee flexion at the beginning of swing phase was tested in the stimulation schedule D. The stimulation schedule E was to test effect of the stimulation schedule D when the stimulation of the soleus substituted for the gastrocnemius medialis in inducing the ankle plantar flexion. Possibility of using stimulation schedule based on EMG data [4] was tested in the stimulation schedule F.

2. 2 Computer Simulation

We designed musculo-skeletal model for hemiplegic FES gait. The model consisted of the electrically stimulated muscle model [5] and the skeletal system model. The skeletal system model consisted of a paralysed leg and a normal leg. The paralysed leg model was activated by the electrically stimulated muscle. Movements of the normal leg model were simulated using the joint angle trajectories measured from a normal subject. Parameters values of musculo-skeletal model were obtained from literature [6].

Computer simulation test of the designed stimulation schedules was performed using a set of fuzzy controllers [3]. In this test, the cycle-to-cycle control was initiated with zero burst durations of stimulation pulses for 200-cycles stimulation courses of swing gait. The burst duration of the vastus muscles at the beginning of control of stimulation (co-activation with the hamstings) of the stimulation schedule D was determined by a ratio of it to the stimulation burst duration of the hamstrings. We tested the stimulation schedule D using five different burst durations of the stimulation of the vastus muscles: 0.1, 0.2, 0.3, 0.4 and 0.5 of the burst duration ratio. The controlled joint angles of each stimulation schedule were obtained under the condition that all the controlled joint angles reached the target joint angles (the well-controlled gait). In case the well-controlled gait could not be reached, the joint angles were obtained in the 200th cycle.

The controlled joint angle trajectories were evaluated by comparing them to the joint angle trajectories of the normal gait. Criteria of evaluation of the stimulation schedules were root-mean-squared (RMS) error, stride length and minimum foot clearance. The RMS error was the RMS value of difference between the controlled joint angle generated in computer simulation and the joint angle measured from normal subjects. The stride length was defined as distance of horizontal displacement of the heel from the beginning of the swing gait to the end of that. The minimum foot clearance was defined as the minimum height of the toe during the swing phase.

3. Results

The trajectories of the controlled joint angles of each stimulation schedule obtained from the computer simulations are shown in Fig.2 comparing to the joint angles measured from the normal

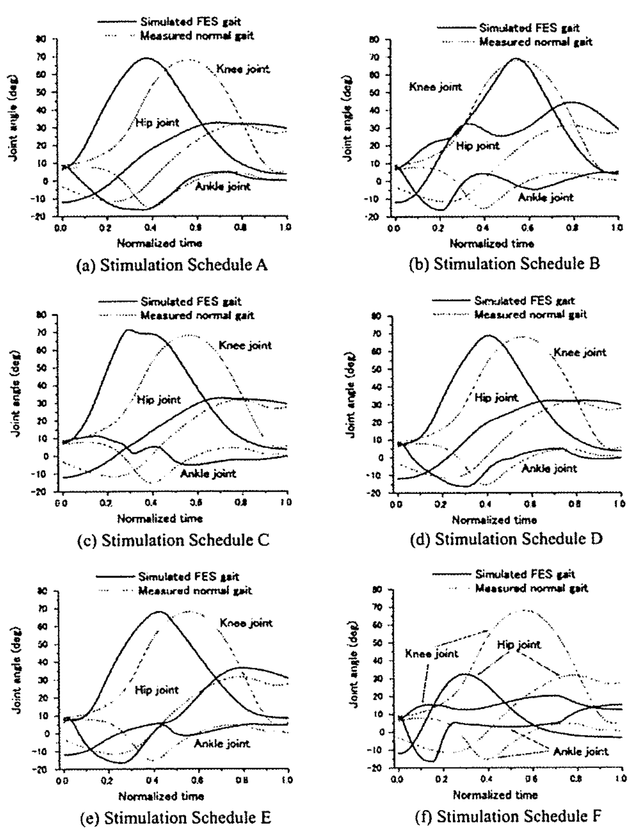

Fig.2 Trajectories of the controlled joint angles of each stimulation schedule generated by model simulation. The measured joint angle trajectories of the normal gait are also shown in each figure. Hip flexion, knee flexion and ankle dorsiflexion are towards positive angles. Hip extension, knee extension and ankle plantar flexion are shown by negative values.

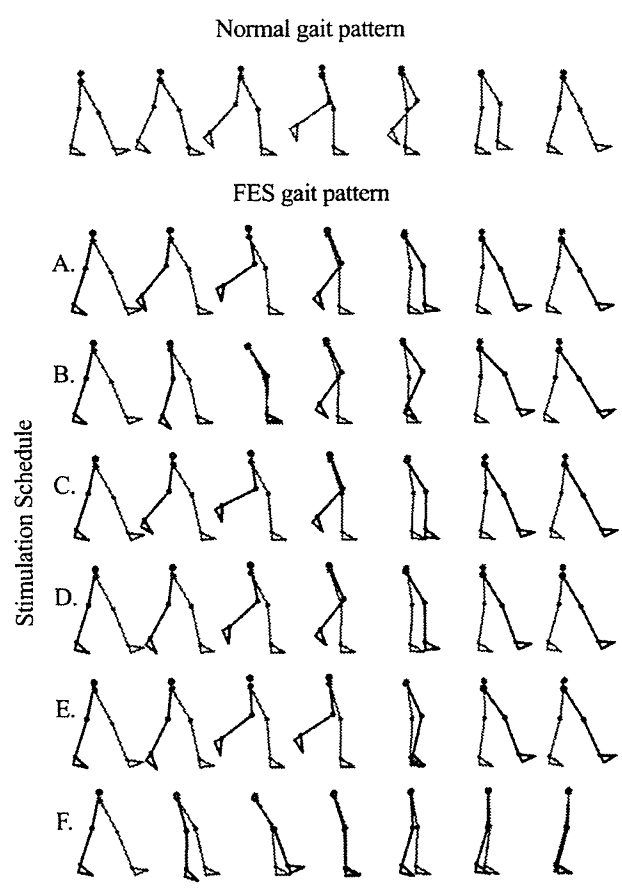

Fig.3 Stick picture of the controlled gait pattern generated by each stimulation schedule. The black leg in the simulated FES gait is the controlled paralyzed swing leg and the grey leg is the normal stance leg.

Table 1 Performance evaluation of stimulation schedule

Stimulation Schedule	RMS error (deg)				Stride length (cm)	Min foot clearance (cm)	Timing of min foot clearance (%)
	Hip angle	Knee angle	Ankle angle	Total			
A	9.7	22.9	7.4	40.0	120.3	2.0	59.4
B	19.3	6.8	11.1	37.2	120.3	-6.1	19.0
C	8.1	24.1	8.0	40.2	120.3	0.1	59.6
D	10.2	19.4	8.6	38.2	120.7	2.2	58.1
E	6.2	16.7	10.3	33.2	119.3	0.3	62.1
F	27.4	28.5	12.2	68.1	67.3	-6.9	13.1

gait. Time was normalized to the duration of the swing gait. Result of performance evaluation of the stimulation schedules was summarized in Table 1. Stick pictures shown in Fig.3 represents pattern of the FES-induced gait of each stimulation schedule and the normal gait.

Stimulation of the hamstrings and the gastrocnemius medialis in the stimulation schedule A caused beginning of knee flexion earlier than the normal gait in the normalized time. However, the stick picture in Fig.3 shows that the controlled gait pattern was not absolutely different from the normal gait pattern. The stimulation schedule B resulted in the gait pattern that was different from the normal gait (Fig.3). The stimulation schedule C caused longer stimulation of the hamstrings resulting in an early flexion of the knee joint as shown in Fig.2 (c). Additionally, the target joint angle of the maximum ankle plantar flexion was not realized as shown in Fig.2 (c) and the minimum foot clearance was very small (Table 1). In case of stimulation schedule D, when the burst duration was greater than 0.3, the number of cycle required to reach the well-controlled gait increased. In case the burst duration ratio was 0.3 (Fig.2 (d)), the RMS error of the knee joint angle was smaller than that of the stimulation schedule A (Table 1). The stimulation of the soleus to induce ankle plantar flexion at the beginning of swing phase of the stimulation schedule E resulted in very small minimum foot clearance (Table 1). The stimulation schedule F resulted in beginning of flexion of the hip joint earlier than the normal gait. The knee could not reach the target of maximum knee flexion angle (Fig.2 (f)). The gait pattern was obviously different from the normal gait pattern (Fig.3).

4. Discussion

Considering the values of evaluation criteria, the stimulation schedule D is preferable to other stimulation schedules. The stimulation schedule A may also be accepted in clinical use. The total RMS errors of both stimulation schedules were not significantly different. The values of minimum foot clearance of the both stimulation schedules were close to the average value of the normal gait (2.19cm)[7]. If the controlled joint angles that are highly similar to the angles of the normal gait are desirable, modulation of stimulation pulse intensity is considered to be an alternative. Although the gait patterns of the stimulation schedules C and E in Fig.3 were not so far from the normal gait pattern, these stimulation schedules should not be used in clinical application because the values of the minimum foot clearance were very small. Control of FES-induced gait is not a single-solution problem. Assessment of the controlled FES gait on the point of view of the quality of the gait performed in this study is one of method in evaluating the design of the stimulation schedule for FES gait control. Two acceptable stimulation schedules were found by the evaluation. Appropriate stimulation schedule for each patient should be determined in clinical tests.

Combination of the information of timing pattern of muscle activation and the knowledge about joint movements and muscle function will be necessary in design of the stimulation schedule for FES gait. Although the stimulation schedule F was based on timing pattern of the EMG, the gait pattern generated by the stimulation schedule F was far from the normal gait pattern. The timing patterns of several muscles can be easily captured from EMG. However, in current state of the cycle-to-cycle control, the electrical stimulation was in fixed intensity. Furthermore, the EMG-based stimulation schedule requires several corrections. Kobetic and Marsolais[8] reported that an initial stimulation pattern based on the EMG pattern could not generate appropriate joint angle trajectories. The stimulation pattern was refined through several manual corrections during experiment. The manual correction of the stimulation pattern during the experiment would be burden to the patient. Although a final stimulation pattern of one patient can be generated, the manual correction is required when it is applied to other patient.

The co-activation of the iliopsoas, the hamstrings and the vastus muscles was found to be effective in controlling swing phase. Although change of timing of the stimulation of the hamstrings in the stimulation schedule B improved the knee joint angle trajectory, the omitted co-activation of the biceps femoris long head and the iliopsoas at the beginning of the swing control caused excessive hip flexion as seen in Fig.2 (b). The co-activation of the vastus muscles and the hamstrings in the stimulation schedule D could improve knee joint angle trajectory as seen in decreasing of the RMS error. The stimulation of the gastrocnemius medialis induced the ankle joint reaching the target of the maximum ankle plantar flexion angle. However, significance of the gastrocnemius stimulation in swing phase control is not clear yet. The results showed that absence of the gastrocnemius medialis stimulation and replacement of the gastrocnemius medialis stimulation with the soleus generated the trajectory of the ankle joint that was not significantly different from the normal gait trajectory. The absence of the gastrocnemius medialis stimulation would not become a severe problem in swing phase control. The co-activation of the tibialis anterior and the soleus was found to be effective in controlling the ankle joint at the end of the swing phase because it increased the ankle joint stiffness. Without the co-activation of the tibialis anterior and the soleus in the stimulation schedules B and F, the ankle joint angle was influenced by the hip and the knee joint movements as seen in Figs 2 (b) and 2 (f), respectively.

Generation of the stimulation schedule for the cycle-to-cycle control based on qualitative knowledge about the joint movements and muscle functions is easier than the generation of stimulation pattern through an optimization method. Yamaguchi and Zajac [9] generated stimulation pattern for FES gait by a dynamic optimization method. The optimized stimulation pattern was tested through a computer simulation. The joint angle trajectories generated by the optimized stimulation pattern was claimed to be similar to the joint angle trajectories of the normal gait. However, because of the complexity of the skeletal system model, the optimization method for generation of the stimulation pattern for FES gait was difficult. Additionally, the optimized stimulation pattern would be impractical for implementing closed-loop control. In the point of view of the simplicity of design and practicality in the closed-loop control scheme, the knowledge based stimulation schedule in this study is preferable.

5. Conclusions

We tested six different stimulation schedules for the cycle-to-cycle control of swing phase of FES-induced hemiplegic gait in the point of view of the quality of gait by computer simulation. The acceptable stimulation schedules were found by considering the evaluation criteria. The results showed that combination of the information of timing pattern of muscle activation and the knowledge about joint movements and muscle function would be necessary in design of the stimulation schedule for FES gait. The co-activation of the iliopsoas, the hamstrings and the vastus muscles at the beginning of swing phase and that of the tibialis anterior and the soleus at the end of swing phase were found to be effective in swing phase control. Future study will be addressed to clinical test of the acceptable stimulation schedules concluded from the present study.

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Development of computer simulation tools for model simulation study on FES control of the upper limb

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1. Introduction

Human movements are produced by contractions of muscles that develop enough forces to move limbs against the external loads. The normal muscle function requires intact connections among the CNS (central nervous system), the spinal cord, and the muscle. Damage of the brain or spinal cord injury (SCI) interrupts the command signal to reach the muscle. People with brain damage or SCI may have loss of motor control resulting in loss of the functional movements in daily life such as standing, locomotion, and reaching.

In a person with paralysis, the loss of command signal from the CNS can be substituted by an artificial electrical stimulation on the peripheral nervous system or the muscle. This electrical stimulation acts in the same way as electrical impulse from the CNS, resulting in muscle contractions and causing movements or sensations. This method is called Functional Electrical Stimulation (FES), and its aim is to provide muscular contraction and produce a functionality of useful movement [1].

FES is an effective method for the restoration of paralyzed motor functions. Current clinical FES system in Japan has adopted open-loop control that uses multichannel stimulation data. The stimulation data are created from electromyogram (EMG) signal measured from neurologically intact subjects. This method is an effective and practical method for clinical applications. The multichannel open-loop control system, however, is desired to be improved in stability and safety of FES induced motions.

The human movements induced by FES require the appropriate control method that can restore the desired functional movements. However, controlling the FES-induced human movements is difficult and complex due to the non-linearity of neuro-muscular system response, variability of response of the stimulated muscle, significant time delay, and muscle fatigue.

Research work on development of FES controller or control method is usually performed with neurologically intact subjects and/or motor disabled patients in order to examine the controller or the control method. However, the burden to the subjects and low reproducibility are common problems of these FES studies.

In our research group, a musculoskeletal model of the upper limb was constructed for the purpose of using in FES research work. Fig.1 shows the outline of the musculoskeletal model. The muscle model consists of the Hill type contractile element and a passive viscoelastic element. Nonlinear characteristics such as recruitment property, length-force relationship, velocity-force relationship and moment arm were described in the model. This model was examined by comparing computer simulation results of closed-loop control of the wrist joint with those on neurologically intact subjects. The simulation results were found to be similar to experimental results in many ways.

The problems on developing FES controller can be partially solved by using a musculoskeletal model. We are developing the computer simulation tool including musculoskeletal model, especially for development of more effective controller for upper limb movement.

2. Problems in current computer simulation study

The musculoskeletal model constructed in our group can be useful tool to study FES control method. However, it is necessary to set parameter values of muscle properties, joint characteristics and experimental conditions for numerical computation. Using the musculoskeletal model, activation torque and viscoelastic torque are calculated, and then the motion equation is solved. Only the numerical data that indicate joint angles as response to stimulation are obtained by the current simulation system. Therefore it is required to plot a graph of joint angles as the function of time or stimulation intensity. Parameters of the musculoskeletal model and FES controller are modified based on simulation results. Then, the computer simulation program is run again. This is inefficient and laborious work.

Additionally, the current simulation system has motion equation for only 4 degrees of freedom of movement. It simulates flexion/extension of the elbow joint, pronation/supination of the forearm, radial/ulnar flexion and palmar/dorsi flexion of the wrist joint. But it makes no account of internal/external rotation of the humerus, flexion/extension and adduction/abduction of the shoulder joint. There is no problem for simulation in case of control of 2 degrees of freedom of movement for wrist joint. However, the motion equation for 7 degrees of freedom of movement is required if it is needed to simulate control of movement including the elbow joint.

4. Development of computer simulation tools

In order to overcome above problems, we improved computer simulation program by including GUI (graphical user interface) and the motion equation for 7 degrees of freedom of movement. This simulation tool can simulate movement of shoulder joint and it doesn't outputs only numerical data, but also graphic views. Fig.2 indicates an example of the screen image of simulation tool. As shown in this figure, the simulation tool solves the motion equation drawing graphs of joint angles as the function of stimulation intensity. Therefore we can get image of joint angle trajectory easily. Additionally, model parameter values can be changed easily by using the developed GUI in real-time. This function makes it easy to perform various conditions of model simulation eliminating repeated trials for manual parameter tuning.

Furthermore, this simulation tool was modified to calculate angles of rotation of the humerus, extension/ flexion and adduction/abduction of the shoulder joint by using motion equation for 7 degrees of freedom of movement. It is very important to study control method of movement of multi degrees of freedom including movement of the elbow joint.

5. Future directions

Using the previous musculoskeletal model, applications of the feedback error learning (FEL) and fuzzy control to FES control have also been discussed [2][3]. These controllers will be examined to develop a controller for 4 degrees of freedom of movements in computer simulations with the developed simulation tool.

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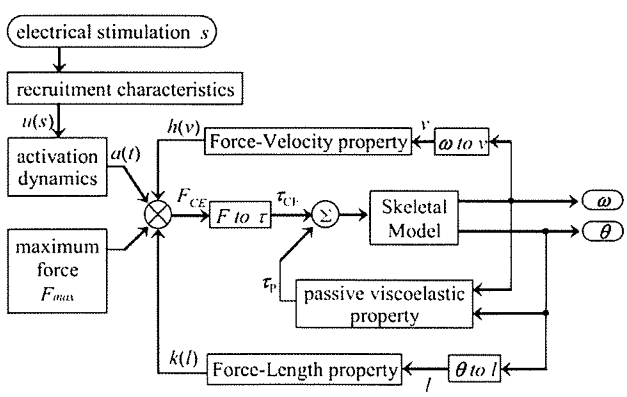

Fig.1 Outline of the musculoskeletal model.

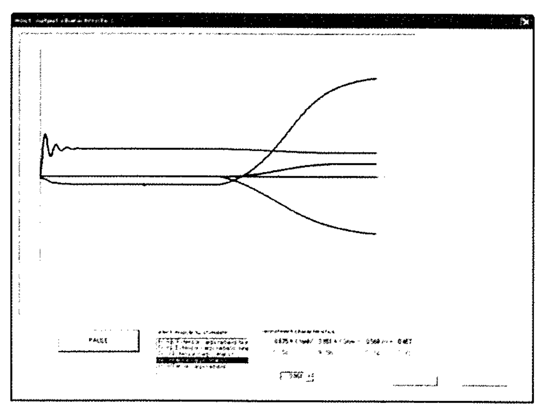

Fig.2 Screen image of new simulation tool

PAPER

Design of Fuzzy Controller of the Cycle-to-Cycle Control for Swing Phase of Hemiplegic Gait Induced by FES

Achmad ARIFIN[†a)], Takashi WATANABE[††], *Nonmembers*, *and* Nozomu HOSHIMIYA[†††], *Fellow*

SUMMARY The goal of this study was to design a practical fuzzy controller of the cycle-to-cycle control for multi-joint movements of swing phase of functional electrical stimulation (FES) induced gait. First, we designed three fuzzy controllers (a fixed fuzzy controller, a fuzzy controller with parameter adjustment based on the gradient descent method, and a fuzzy controller with parameter adjustment based on a fuzzy model) and two PID controllers (a fixed PID and an adaptive PID controllers) for controlling two-joint (knee and ankle) movements. Control capabilities of the designed controllers were tested in automatic generation of stimulation burst duration and in compensation of muscle fatigue through computer simulations using a musculo-skeletal model. The fuzzy controllers showed better responses than the PID controllers in the both control capabilities. The parameter adjustment based on the fuzzy model was shown to be effective when oscillating response was caused due to the inter-subject variability. Based on these results, we designed the fuzzy controller with the parameter adjustment realized using the fuzzy model for controlling three-joint (hip, knee, and ankle) movements. The controlled gait pattern obtained by computer simulation was not significantly different from the normal gait pattern and it could be qualitatively accepted in clinical FES gait control. The fuzzy controller designed for the cycle-to-cycle control for multi-joint movements during the swing phase of the FES gait was expected to be examined clinically.

key words: *cycle-to-cycle control, FES-induced gait, multi-joint control, fuzzy controller, adaptive fuzzy controller*

1. Introduction

Functional electrical stimulation (FES) has been utilized to restore gait in patient with impairment of the central nervous system caused by the spinal cord injury or the stroke [1], [2]. The human gait is a complex task, therefore FES-induced gait requires appropriate control methods. The control methods used in FES involve open-loop and closed-loop controls. The previous experimental researches [3], [4] showed limitation of the open-loop control to compensate muscle fatigue. A trajectory-based closed-loop control for the knee joint angle of paraplegic gait had poor tracking and oscillating responses, and could not reach full knee extension angle in some trials [5]. Capability of tracking a target joint angle is still not clear.

The cycle-to-cycle control was expected to be an alternative to the trajectory-based closed-loop FES gait control. Its capability to realize the target joint angle had been shown in experimental test of controlling maximum knee extension angle [6] or hip joint angle range [7] using a proportional-integral-derivative (PID) controller. However, these experimental tests were performed in single-joint control. The cycle-to-cycle control should be implemented for controlling multi-joint movements in clinical use. Because there is a lack of method of determination of parameter values of the PID controller of the cycle-to-cycle control, implementation of the cycle-to-cycle control for multi-joint movements using the PID controller is difficult.

We proposed fuzzy controller to realize the cycle-to-cycle control. The fuzzy controller was shown to be superior to the PID controller in controlling the knee joint movements of swing phase of FES gait [8]. The fuzzy control scheme had been shown to be effective in developing nonlinear controllers [9]. Using the fuzzy controller, system identification to determine the controller parameter values could be eliminated and design procedure was simplified. In order to develop a practical cycle-to-cycle control, the fuzzy controller has to be realized for controlling the multi-joint movements.

In this paper, we designed the fuzzy controller of the cycle-to-cycle control for multi-joint movements of the swing phase of FES gait. First, we designed three different fuzzy controllers (a fixed fuzzy controller, a fuzzy controller with parameter adjustment based on gradient descent method, and a fuzzy controller with parameter adjustment based on a fuzzy model) and two PID controllers (a fixed PID controller and an adaptive PID controller) for controlling two-joint (knee and ankle) movements. They were examined through computer simulations with the aim of exploring the control capability of the designed controllers in the basic structure of the lower limbs. Based on these results, we realized the cycle-to-cycle control for three-joint (hip, knee, and ankle) movements. The swing phase with the three-joint movements controlled by the designed controllers was evaluated in the computer simulations comparing to the measured joint angle trajectories of the normal gait.

2. Framework of the Cycle-to-Cycle Control

Gait is one of the cyclic movements. Each gait cycle is divided into the stance phase and the swing phase. In a cer-

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tain sub-phase of the gait, the joint movements reach certain joint angles (e.g., maximum knee flexion angle of swing phase, maximum ankle dorsifelxion angle of swing phase, ankle joint angle at initial contact). In the cycle-to-cycle control, stimulation burst duration, TB, is regulated, while pulse amplitude, pulse width, and frequency of stimulation pulse are fixed. Each muscle is stimulated by single burst of electrical pulses to produce relevant joint movement reaching a target joint angle. The controlled joint angle of the previous cycle is fedback to the controller. Regulation of the stimulation burst duration of the current cycle of gait is based on difference between the target joint angle and the controlled joint angle of the previous cycle.

A set of certain target joint angles and a stimulation schedule are required to realize the cycle-to-cycle control. We performed an experiment to measure the hip, knee and ankle joint angles during level gait. Five healthy subjects (males, 24±2.9 years old, 170.2±4.5 cm, 60.6±5.4 kg) participated in the experiment. Purpose of the experiment was explained to each subject and subject's consent was obtained. Three goniometers (M180, Penny & Giles, UK) were used to measure the hip, knee, and ankle joint angles of right leg of the subject. Two force sensitive resistors (FSRs) were attached to the heel and the toe of the foot to detect the stance and the swing phases. The subjects were instructed to perform five trials of level gait with comfortable speed. A trial of gait consisted of seven gait cycles. The output signals of the goniometers and the FSRs were recorded into the data cartridge using RTD-145 (TEAC, Japan) and digitized by an A/D converter (AT-MIO-16E-10, National Instruments) with 1 kHz of sampling frequency. Data processing was performed using MATLAB version 6.5 Release 13. The data of FSRs were low-pass filtered using fourth order, zero phase lag, Butterworth digital filter with cut-off frequency of 1 Hz. Data of the joint angles were smoothed using the fourth order, the zero phase lag, the Butterworth digital filter with 20 Hz of cut-off frequency. Gait analysis of the data collected from the experiment was performed to determine the target joint angles. Parameter $\Delta\theta$ was also determined in order to evaluate whether the target joint angle was reached or not. The controlled joint angle reached the target when the absolute error was less than or equal to $\Delta\theta$ as illustrated in Fig. 1. Values of $\Delta\theta$ of each controlled joint angle was set

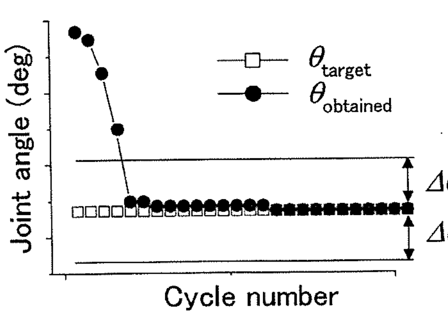

Fig. 1 Illustration of $\Delta\theta$ in control evaluation. $\theta_{obtained}$ is the obtained joint angle in each cycle of gait and θ_{target} is the target joint angle.

by average value of intra-subject standard deviation of each target joint angle from the gait analysis. Values of the target joint angles and the $\Delta\theta$ are shown in Table 1.

We developed the stimulation schedule to develope the joint movements of the swing phase as shown in Fig. 2, which was designed based on the knowledge about the joint movements during swing phase and muscle functions to produce the relevant joint movements [10]. Stimulations of the iliopsoas, the biceps femoris short head (BFSH) and the biceps femoris long head (BFLH), the vastus muscles and the rectus femoris, the gastrocnemius medialis, and the tibialis anterior were controlled to induce the joint movements reaching the following target joint angles: maximum hip flexion angle, maximum knee flexion angle, maximum knee extension angle, maximum ankle plantar flexion angle, and maximum ankle dorsiflexion angle, respectively. After the hip joint reached the target maximum hip flexion angle, the iliopsoas was stimulated again to keep hip flexion and to reach the target of hip joint angle at initial contact. The tibialis anterior and the soleus were stimulated simultaneously to reach the target of ankle joint angle at initial contact. Beginnings of the muscle stimulation were at the maximum hip extension, maximum knee extension and maximum an-

Table 1 Target joint angle and $\Delta\theta$.

Joint	Angle	Target	$\Delta\theta$
Hip	max. flexion	32.4°	2.0°
	initial contact	29.3°	2.8°
Knee	max. flexion	69.0°	1.9°
	max. extension	3.6°	2.7°
Ankle	max. plantar flexion	−16.4°	3.4°
	max. dorsiflexion	4.9°	1.3°
	initial contact	−0.3°	1.3°

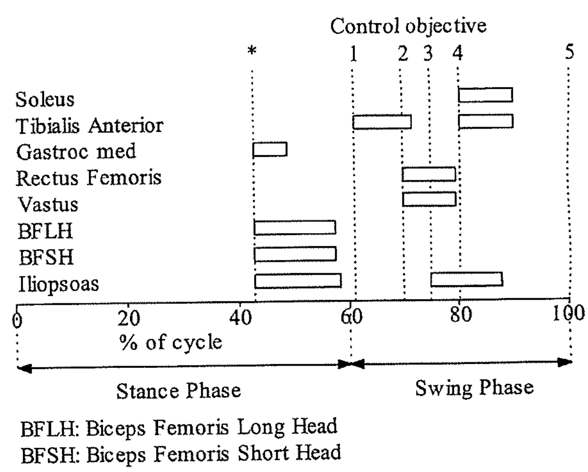

Fig. 2 The stimulation schedule. *: the beginning of the stimulation (the maximum hip extension angle, the maximum knee extension angle of the end of the stance phase, and the maximum ankle dorsiflexion angle of the end of the stance phase). The control objective: 1: the maximum ankle plantar flexion angle, 2: the maximum knee flexion angle, 3: the maximum hip flexion angle, 4: the maximum ankle dorsiflexion angle, and 5: the maximum knee extension angle and the hip and the ankle angles at initial contact.

kle dorsiflexion angles at the end of stance phase. In our result of the gait analysis, timing of those maximum joint angles in a cycle of gait varied among subjects. In order to facilitate the computer simulation, we assumed these maximum joint angles were occurred simultaneously.

3. Musculo-Skeletal Model for FES Gait

We developed the electrically stimulated musculo-skeletal model for FES gait based on Eom et al. [11]. Values of the musculo-skeletal model parameters were obtained from Ogihara et al. [12]. The joint movements were considered as the movements of a multiple pendulum. The motion equation was derived from the model using the Lagrange function as shown in Eq. (1),

$$\mathbf{M}\ddot{\boldsymbol{\theta}} + \mathbf{C}\dot{\boldsymbol{\theta}} + \mathbf{G} = \boldsymbol{\tau} \qquad (1)$$

where $\boldsymbol{\theta}$, $\dot{\boldsymbol{\theta}}$, $\ddot{\boldsymbol{\theta}}$ are vectors of the joint angle, the joint angular velocity, and the joint angular acceleration, $\mathbf{M}$ is the inertial matrix, $\mathbf{C}$ is the coriolis vector, $\mathbf{G}$ is the gravitational vector, and $\boldsymbol{\tau}$ is vector of the joint torque. The model for the three-joint movements consisted of the paralyzed and the normal legs. In this study, the joint angles of the normal leg were simulated using the values of joint angles measured from the normal subject. The joint movements developed by the electrical stimulation of the muscles were calculated by integrating the motion equation in Eq. (1) by the fourth order Runge-Kutta method with $10\,\mu$s integration time step.

In order to test the capabilities of the designed controllers in controlling different subjects, we created twenty different subject models. The twenty different subject models were expressed by changing values of maximum muscle forces (50%–150%), mass of the thigh, the shank, and the foot (50%–150%), and/or length of the thigh, the shank, and the foot (75%–125%) of a reference subject model.

4. Computer Simulation Test in Two-Joint Control

The stimulated muscles in the two-joint control were the BFSH, the vastus muscles, the gastorocnemius medialis, the tibialis anterior, and the soleus. The sequence of the stimulation of the muscles was based on the stimulation schedule shown in Fig. 2. We designed the three different fuzzy controllers, controller A was a fixed parameter fuzzy controller, controller B was a fuzzy controller with the gradient descent based parameter adjustment and controller C was a fuzzy controller with fuzzy model parameter adjustment and oscillation detection. A PID controller (controller D) and an adaptive PID controller (controller E) were also designed. Concrete descriptions of the design of the fuzzy and the PID controllers are given in the following subsections.

4.1 Design of Fuzzy Controller

Algorithm of regulation of the stimulation burst duration, TB, by the fuzzy controller is shown in Eq. (2),

$$TB[n] = TB[n-1] + \Delta TB[n] \qquad (2)$$

where $TB[n]$ is the stimulation burst duration for the current cycle, $TB[n-1]$ is the stimulation burst duration of the previous cycle, and $\Delta TB[n]$ is the output of the fuzzy controller. The fuzzy controllers were designed through five design steps: (i) selection of controller inputs, (ii) assignment of the fuzzy membership functions, (iii) design of the fuzzy rules, (iv) assignment of fuzzy inference and deffuzification, and (v) design of parameter adjustment.

4.1.1 Selection of Controller Inputs

The inputs for the BFSH, the gastrocnemius medialis, and the iliopsoas controllers were the *errors*. These controllers were designed as single-input single-output (SISO) controllers. The *error* was defined as the difference between the target and measured joint angles. The controllers for the vastus muscels, the soleus, and the tibialis anterior were designed as multi-input single-output (MISO) controllers. The inputs of these controllers were *error* and *desired range* of joint angle. The desired range of the knee extension angle was defined as the difference between the obtained maximum knee flexion angle of the current cycle and the target maximum knee extension angle. The desired range of the ankle dorsiflexion angle was defined as the difference between the obtained maximum ankle plantar flexion angle of the current cycle and the target maximum ankle dorsiflexion angle. The desired range of the ankle angle at the initial contact was defined as the difference between the obtained maximum ankle dorsiflexion of the current cycle and the target ankle joint angle at the initial contact.

4.1.2 Assignment of Fuzzy Membership Functions

Input membership functions were expressed as triangular and trapezoidal fuzzy sets. Output membership function was expressed in the fuzzy singletons. The fuzzy linguistic terms of the input and output variables are shown in Table 2. The values of parameters of the fuzzy membership functions were assigned and refined through preliminary computer simulation.

Table 2 Fuzzy linguistic term.

Input		Output	
desired range	*error*	ΔTB^*	
S	NL	NL2	S: small
M	NM	NL1	M: medium
L	NS	NL	L: large
	Z	NM	NL2: negative large 2
	PS	NS	NL1: negative large 1
	PM	Z	NL: negative large
	PL	PS	NM: negative medium
		PM	NS: negative small
		PL	Z: zero
		PL1	PS: positive small
		PL2	PM: positive medium
			PL: positive large
			PL1: positive large 1
			PL2: positive large 2