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Evaluation of the influence of muscle deactivation on other muscles and joints during gait motion

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Abstract

When any muscle in the human musculoskeletal system is damaged, other muscles and ligaments tend to compensate for the role of the damaged muscle by exerting extra effort. It is beneficial to clarify how the roles of the damaged muscles are compensated by other parts of the musculoskeletal system from the following points of view: From a clinical point of view, it will be possible to know how the abnormal muscle and joint forces caused by the acute compensations lead to further physical damage to the musculoskeletal system. From the viewpoint of rehabilitation, it will be possible to know how the role of the damaged muscle can be compensated by extra training of the other muscles. A method to evaluate the influence of muscle deactivation on other muscles and joints is proposed in this report. Methodology based on inverse dynamics and static optimization, which is applicable to arbitrary motion was used in this study. The evaluation method was applied to gait motion to obtain matrices representing (1) the dependence of muscle force compensation and (2) the change to bone-on-bone contact forces. These matrices make it possible to evaluate the effects of deactivation of one of the muscles of the musculoskeletal system on the forces exerted by other muscles as well as the change to the bone-on-bone forces when the musculoskeletal system is performing the same motion. Through observation of this matrix, it was found that deactivation of a muscle often results in increment/decrement of force developed by muscles with completely different primary functions and bone-on-bone contact force in different parts of the body. For example, deactivation of the iliopsoas leads to a large reduction in force by the soleus. The results suggest that acute deactivation of a muscle can result in damage to another part of the body. The results also suggest that the whole musculoskeletal system must go through extra retraining in the case of damage to certain muscles.
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1. Introduction

When a muscle of the musculoskeletal system is damaged, i.e., its force development capacity is impaired (Shields, 2002; van Kuijk et al., 2002), it is usually possible to either (1) compensate for the damage to the muscle by modifying (usually enhancing) the efforts of other muscles while still performing the same (or very similar) motion as a whole (Ageberg, 2002; Liu and Maitland, 2000) or (2) to drastically modify the motion into another motion that naturally results in a lower mechanical load being imposed on the damaged muscle (Hooper et al., 2002; Sudarsky, 2001). The first option is usually preferred to the second option for aesthetical, social, physiological and biomechanical reasons. When the first option is chosen by the patient and clinician, the patient performs the same (or very similar) motion with different patterns of muscular force development by other muscles, which results in different profiles of bone-on-bone contact forces being developed at the joints (Winter, 1990). Therefore, when choosing the first option, it is important to quantitatively evaluate the impact of the damage to a muscle on other components of the musculoskeletal system in order to avoid causing additional physical problems (Herzog et al., 1998; Suter and Herzog, 2000) and also to know how much extra training by the other muscles is needed to compensate for the role of the damaged muscle.

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Many researchers have analyzed the contribution of some specific muscles to performance by using optimization methods based on forward dynamics (Nigg and Herzog, 1999). For example, Pandy and Zajac (1991) analyzed the role of biarticular muscles in maximal jumping, and they concluded that biarticular muscles contribute to jumping performance by redistributing segmental energy within the musculoskeletal system but without generating energy by themselves (a conclusion that is consistent with that made by Jacobs et al. (1996)). Neptune et al. (2001) analyzed the role of the plantarflexor muscles during gait, and they calculated the degree to which these muscles contribute to propelling the trunk in the forward direction (induced acceleration; Yamaguchi, 2001). Piazza and Delp (1996) examined the contribution of muscle forces to knee flexion during the swing phase of normal gait.

However, the effects of muscle deactivation, i.e., how other muscles work to compensate for the role of the damaged muscle whose force development capacity has been impaired, were not examined in these studies. In a clinical situation in which a patient chooses to perform the same motion in spite of damage to a muscle (Ageberg, 2002; Liu and Maitland, 2000), it is important to evaluate the impact of deactivation of the muscle on the biomechanics of the whole musculoskeletal system.

The impact of muscle deactivation on the biomechanics of the musculoskeletal system can be evaluated by the use of inverse dynamics, by which the moments developed around joints can be calculated from the kinematic motion as an input. The human musculoskeletal system is endowed with a degree of redundancy, i.e., the number of muscles is larger than the number of degrees of freedom of the system (Yamaguchi, 2001), so that an almost infinite number of combinations of muscle forces are possible in order to achieve a combination of joint moments. Methodologies to address this redundancy aspect have been developed in many studies primarily through optimization approaches such as minimizing the level of muscle activation (Anderson and Pandy, 2001), minimizing muscle stress (Crowninshield and Brand, 1981), and the pseudo-inverse method (Yamaguchi et al., 1995). It has been reported that methods based on inverse dynamics yield a highly realistic output for motions such as human gait (Anderson and Pandy, 2001).

In this paper, we describe an inverse dynamics method that enables quantification of the dependence between deactivation of a muscle and changes in the muscular effort of other muscles and bone-on-bone forces while performing the same movement to yield the same kinematics. Although this methodology is applicable to practically any type of motion, gait motion was chosen as an example in this study.

2. Methods

The musculoskeletal model developed by Delp (1990) was used in this study. Data used in the model include the attachment sites of 43 muscles on each leg and physiological parameters such as the lengths of tendons and muscular filaments. The lower half of the body (Fig. 4) is composed of the pelvis and the femur, tibia, patella, talus, calcaneous and toes in each leg. The joints of the legs are assumed to be either 3-DOF gimbal joints (hip joints) or 1-DOF joints (knee, ankle, calcaneous, and metatarsal joints). Therefore, the total number of degrees of freedom is 14.

Each musculotendon is based on the musculotendon model of Hill (1938), and parameter values were derived from Delp (1990).The musculotendon model is composed of three elements (Fig. 1): a contractile element (CE, representing all the muscle fibers), a parallel elastic element (PEE, representing all connective tissues around the muscles fibers), and a series elastic element (SEE, representing all series of elasticity, including tendons). At each time step, the musculotendon length was determined from the posture (i.e., as a function of joint angles). Thereafter, the range of force developed by a muscle can be calculated at each time step by

$$F_m^{\min} \leqslant F_m(t) \leqslant F_m^{\max}. \tag{1}$$

Here, $F_m(t)$ is the musculotendon force of the mth muscle, and $F_m^{\min}$ and $F_m^{\max}$ are the minimum and maximum forces developed by this muscle, which are determined by their force–length–velocity properties:

$$F_m^{\min} = f(F_m^0, l_m, \upsilon_m, 0),$$
$$F_m^{\max} = (F_m^0, l_m, \upsilon_m, 1),$$

where function $f(F_m^0, l_m, \upsilon_m, a_m)$ is the force–length–velocity surface assumed in the musculoskeletal model (Zajac, 1989), l^m is the length and υ^m is the velocity of the shortening muscle m, and the 0 and 1 in these equations are a_m, the activation level of the muscle that

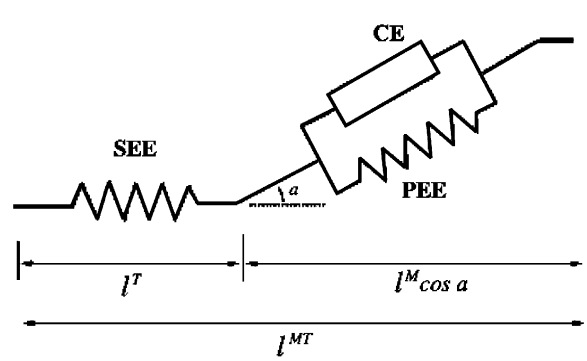

Fig. 1. Muscle model by Hill used in this study. The muscle–tendon complex model consists of three elements: a CE, a SEE, and a PEE. The total length of the muscle–tendon complex, tendon length, muscle fiber length, and pennation angle are represented by l^T, l^M, l^{MT}, and a.

determines the amount of force exerted by the CE ($0 \leqslant a_m \leqslant 1$). The torque developed at each joint was calculated by using inverse dynamics, with kinematic data as input data (Winter, 1990). Torque $\tau_j(t)$ developed at joint j is theoretically generated as follows by the muscles crossing the joint:

$$\tau_j(t) = \sum_m F_m(t) r_{m,j}, \quad j = 1, \ldots, n_{\mathrm{dof}}, \tag{2}$$

where $r_{m,j}$ is the moment arm of muscle m about the jth joint axis, and n_{dof} is the number of degrees of freedom whose torque is assumed here to be generated only by the muscles. They include flexion/extension, adduction/abduction and rotation at the hip, flexion/extension at the knee, and plantarflexion/dorsiflexion at the ankle, and, therefore, by taking into account both legs, $n_{\mathrm{dof}} = 10$.

The muscle forces at each time step were calculated by minimizing the muscle stress (Crowninshield and Brand, 1981):

$$J = \sum_m^{n_m} \left(\frac{F_m(t_i)}{F_m^0} \right) 3, \tag{3}$$

where n_m is the total number of muscles ($n_m = 43 \times 2 = 86$), and F_m^0 is the maximal force parameter of muscle m, which is calculated by the physiological cross-sectional area of the muscle. J was minimized using sequential quadratic programming (Lawrence et al., 1997), which is an optimization method that can minimize an objective function while satisfying equality and inequality constraints. In summary, the muscle forces were calculated by minimizing Eq. (3) while using Eqs. (1) and (2) as constraints.

Once the muscle forces had been obtained, the bone-on-bone forces at the joints were calculated by using the following equation:

$$\boldsymbol{f}_j^r = \sum\nolimits_{m_c} \boldsymbol{F}_{m_c} - \boldsymbol{f}^j, \tag{4}$$

where F_{m_c} is the force developed by the muscles that cross a joint, $\boldsymbol{f}^j$ is the segmental joint reaction force that is calculated by using inverse dynamics (Winter, 1990), and $\boldsymbol{f}_j^r$ is the joint bone-on-bone force.

Injury to or paralysis of a muscle often affects the maximal force development capacity of that muscle (Shields, 2002; van Kuijk et al., 2002). As the force development capacity of a muscle decreases, the patient tries to achieve the same kinematic motion by modifying (usually by enhancing) the efforts of other muscles. In this study, the modification required to compensate for the damage to a muscle was evaluated by calculating the finite difference of the force developed by other muscles to achieve the motion due to the decreased maximal force development capacity of the impaired muscle.

First, a function that determines the muscle force using the maximal force parameters by the method mentioned above is defined here by

$$F_m = \tilde{F}_m(F_0^0, \ldots, F_{n_m-1}^0) \, (m = 0, \ldots, n_m - 1). \tag{5}$$

How muscle m compensates for the deactivated muscle can be calculated by decreasing the amount of the maximal force development capacity:

$$F_m + \delta F_m^i = \tilde{F}_m(F_0^0, \ldots, F_i^0 - \delta F_i^0, \ldots, F_{n_m-1}^0),$$

where δF_i^0 is the decrement of the maximal force development capacity by muscle i, and δF_m^i is the increment/decrement of the force by muscle m. The finite difference that represents the dependency between maximal force development capacity of muscle i and force by muscle m can be obtained by

$$\delta F_m^i = \tilde{F}_m(F_0^0, \ldots, F_i^0 - \delta F_i^0, \ldots, F_{n_m-1}^0) - F_m. \tag{6}$$

The influence of the deactivated muscle on joint bone-on-bone forces can be evaluated by first defining a function that determines the relationship between F_i^0 and the joint bone-on-bone force $\boldsymbol{f}_j^r$ by

$$\boldsymbol{f}_j^r = \tilde{\boldsymbol{f}}_j^r(F_0^0, \ldots, F_{n_m-1}^0). \tag{7}$$

The finite differential value that represents the dependency between maximal force development capacity and joint bone-on-bone force can be calculated as well by

$$\begin{aligned} \delta \boldsymbol{f}_{i,j}^r = & \tilde{\boldsymbol{f}}_j^r(F_0^0, \ldots, F_i^0 - \delta F_i^0, \ldots, F_{n_m-1}^0) \\ & - \tilde{\boldsymbol{f}}_j^r(F_0^0, \ldots, F_{n_m-1}^0), \end{aligned} \tag{8}$$

where $\delta \boldsymbol{f}_{i,j}^r$ is the estimated decrease in the joint bone-on-bone force at joint j when the decrement of the maximal force of muscle i is δF_i^0.

The decrement of the maximal force development capacity was set as $\delta F_i^0 = F_i^0$ to observe the influence of full deactivation of the muscle on other muscles and joints. In some cases, i.e., in cases of deactivation of muscles, such as the tibialis anterior (TIBANT), that play a major role in generation of specific joint torques, no solution for muscle force due to deactivation of the muscle was found. In such case, instead of fully deactivating the muscle, the amount of δF_i^0 was gradually decreased until a solution was found. Then the increase in force by other muscles, δF_m^i, were rescaled in order to estimate the muscle force at full deactivation:

$$\delta F_m^{i\prime} = \frac{F_i^0}{\delta F_i^0} \delta F_m^i. \tag{9}$$

The bone-on-bone contact force was also rescaled as well to observe the effect of full deactivation of each muscle:

$$\delta \boldsymbol{f}_{i,j}^{r\prime} = \frac{F_i^0}{\delta F_i^0} \delta \boldsymbol{f}_{i,j}^r. \tag{10}$$

A 50-year-old male subject with body weight of 51 kg and height of 164 cm performed a normal gait motion

during which kinematic data were recorded using a VICON 512 apparatus (Vicon Motion Systems, Oxford, United Kingdom). The joint angles, i.e., hip flexion/extension, hip abduction/adduction, hip internal/external rotation, knee flexion/extension, and ankle dorsiflexion/plantarflexion angles, were determined. This data collection procedure has been reviewed and approved by the institutional review board (IRB) of RIKEN (Wako-Shi, Saitama, Japan). The procedure for analysis described previously was applied to these kinematic data. The activities of the following muscles were analyzed: m. gluteus medialis (GMED), m. gluteus minimus (GMIN), m. hamstrings (HAM), m. adductor longus, m. adductor magnus, m. tensor fasciae latae, m. gluteus maximus (GMAX), m. iliapsoas (ILPSO), m. rectus femoris (RF), m. vastus (VAS), m. gastrocnemius (GAS), m. soleus (SOL), and m. tibialis anterior TIBANT.

Our method was applied to the support phase of the gait motion. By minimizing Eq. (3), the forces exerted by the leg muscles and the bone-on-bone force of the support leg were calculated. The size of each body segment was linearly scaled to adapt the musculoskeletal model to the subject (Fig. 2). Muscle parameters such as the origins and insertions of muscles and the lengths of fibers and tendons were also scaled. The new inertia and mass of the body segments were each calculated by using these scaling factors.

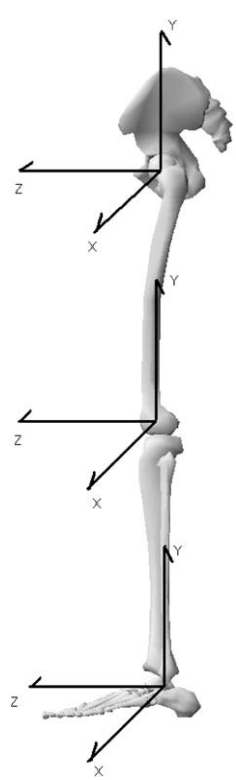

Fig. 2. Local coordinate system of each body segment.

The co-dependence of the muscles was evaluated by deactivating the muscles one by one and then calculating the integral of the definite difference of the force exerted by the other muscles during the motion:

$$\frac{\int_{t_0}^{t_f} \delta F_m^i \, \mathrm{d}t}{t_f - t_0}, \tag{11}$$

where t_0 and t_f are the time of the beginning and the time of the ending of the motion. The integral is divided by the amount of time for normalization. In cases in which muscle i could not be fully deactivated for performing the motion, $\delta F_m^{i\,\prime}$ in Eq. (9) was used instead.

The dependence of bone-on-bone force on muscle force was also evaluated by calculating the integral of the norm of the change in the force during the motion:

$$\frac{\int \left| \delta \boldsymbol{f}_j^i \right| \mathrm{d}t}{t_f - t_0}. \tag{12}$$

$\delta \boldsymbol{f}_{i,j}^r$ in Eq. (10) was used as well in cases muscle i could not be fully deactivated.

3. Results

The forces exerted by the leg muscles were calculated by minimizing Eq. (3) as shown in Fig. 3. The real human muscle force data estimated by the EMG pattern of muscles during normal gait from Perry (1992) are listed together. This estimation was done by multiplying the maximum force parameter F_m^0 from Delp (1990) by the activation data listed in the report by Perry (1992), which were calculated by manual muscle tests. The motion data used in the present study are for half of the gait cycle. Initial contact of the leg started at 11% in this motion data, and toe off started at 95%.

We next calculated the finite difference in muscle forces by changing the maximal force of each muscle. The forces exerted by the leg muscles for the gait motion were calculated by minimizing Eq. (3). The time-course characteristics for the newly estimated muscle forces calculated by deactivating the RF, GMAX, and the HAM are shown in Fig. 4. Examination of Fig. 4 enables the roles of other muscles in augmentation of the deactivated muscle to be identified. For example, when the RF is deactivated, the knee moment that was produced by the RF is provided by the VAS (Fig. 4(d) 30–100%).

When the GMAX was deactivated, the hip extension moment that was produced by the GMAX was augmented by the GMED and HAM (Figs. 4(e), (f) 10–50%) and the additional knee extension moment that was produced by the HAM was counteracted by the VAS (Fig. 4(d), 10–30%).

The increase/decrease in the bone-on-bone force at each joint was also calculated, and the characteristics are

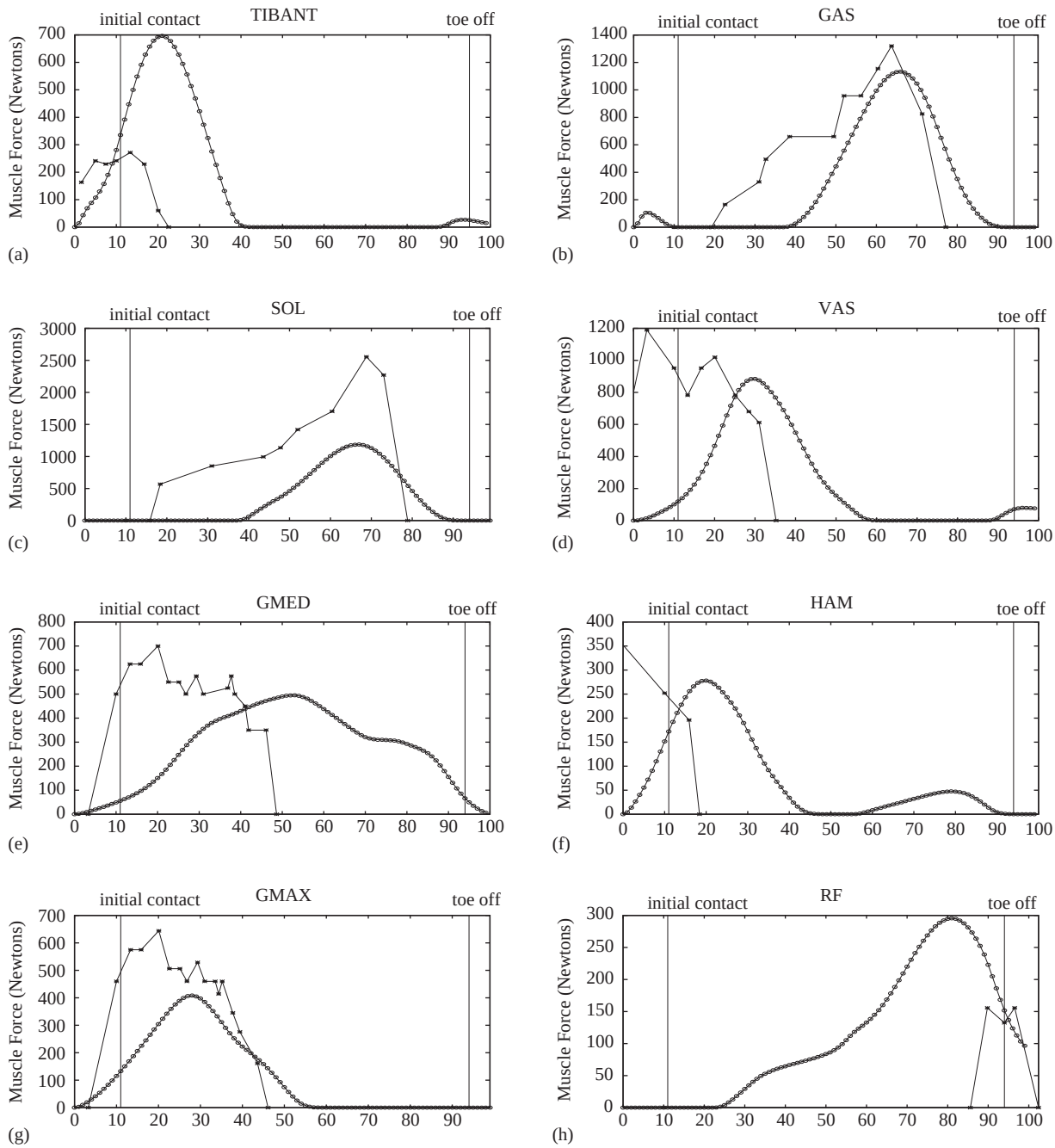

Fig. 3. Muscle force characteristics of the support leg during gait (○) and the corresponding EMG data (*) from Perry (1992). The figures show the characteristics of (a) TIBANT, (b) GAS, (c) SOL, (d) VAS, (e) GMED, (f) HAM, (g) GMAX, and (h) RF.

shown in Fig. 5 (hip joint), Fig. 6 (knee joint), and Fig. 7 (ankle joint).

It can be seen that a change in the maximal muscle force affects not only the joints that it crosses but also all of the joints of the leg. For example, the force at the knee was affected by the GMAX that only crosses the hip, and the force at the ankle was affected by the RF that only crosses the hip and knee.

It should be noted that deactivation of the GMAX mainly affected the bone-on-bone force during the first half of the support phase, while deactivation of the RF affected the latter half of the support phase. In addition, while deactivation of the RF decreased the vertical component of the bone-on-bone force at the knee, it increased the bone-on-bone force at the hip. It is interesting that deactivation of the GMAX affects the bone-on-bone force at the knee more than that at the hip.

The results of calculation of the dependence of muscle forces on each other using Eq. (11) are shown in Table 1,

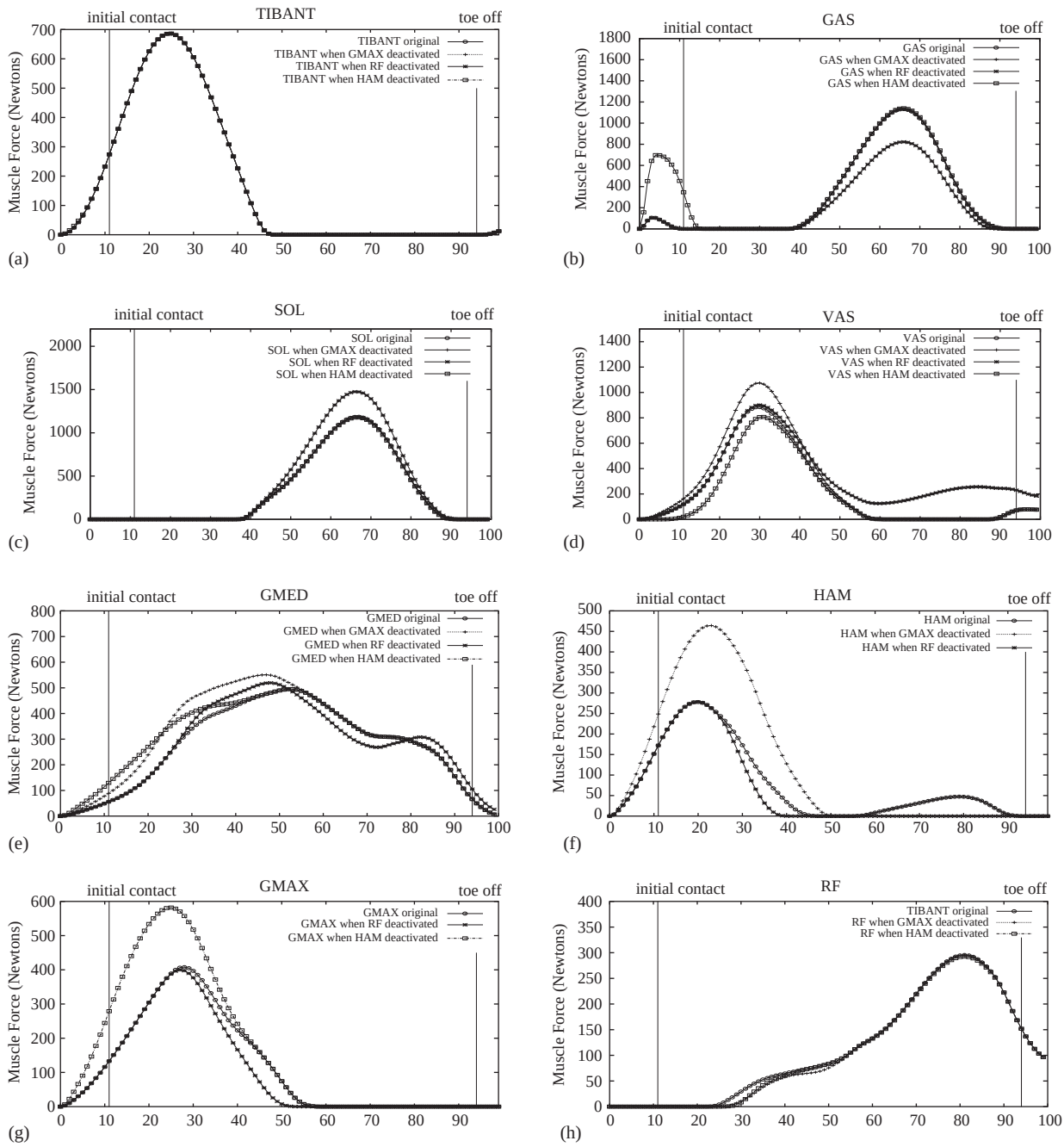

Fig. 4. Original muscle force characteristics (shown by ○) and those after deactivation of the GMAX (+), RF (*), and HAMs (□) muscles. The figures show the characteristics of (a) TIBANT, (b) GAS, (c) SOL, (d) VAS, (e) GMED, (f) HAM, (g) GMAX, and (h) RF.

and relationships between muscle force and bone-on-bone joint force calculated by Eq. (12) are shown in Table 2. The data presented in Table 1 indicate that although muscles that cross the same joints tend to have stronger dependence on each other, some muscles affect the forces of muscles with quite different functions. For example, deactivation of muscles such as the ILPSO and RF affects the force exerted by the SOL, and deactivation of the GAS increases the muscle force exerted by the GMIN and ILPSO. The data presented in Table 2 indicate that deactivation of muscles around the hip joint (such as the HAM and ILPSO) can greatly change the bone-on-bone force at the ankle. Biarticular muscles such as the HAM, GAS, and RF tend to have a strong influence on all muscles and joints of the leg.

4. Discussion

A comparison of the muscle forces calculated by using the static optimization method and the EMG data shows that the shapes of the time–sequence curves are

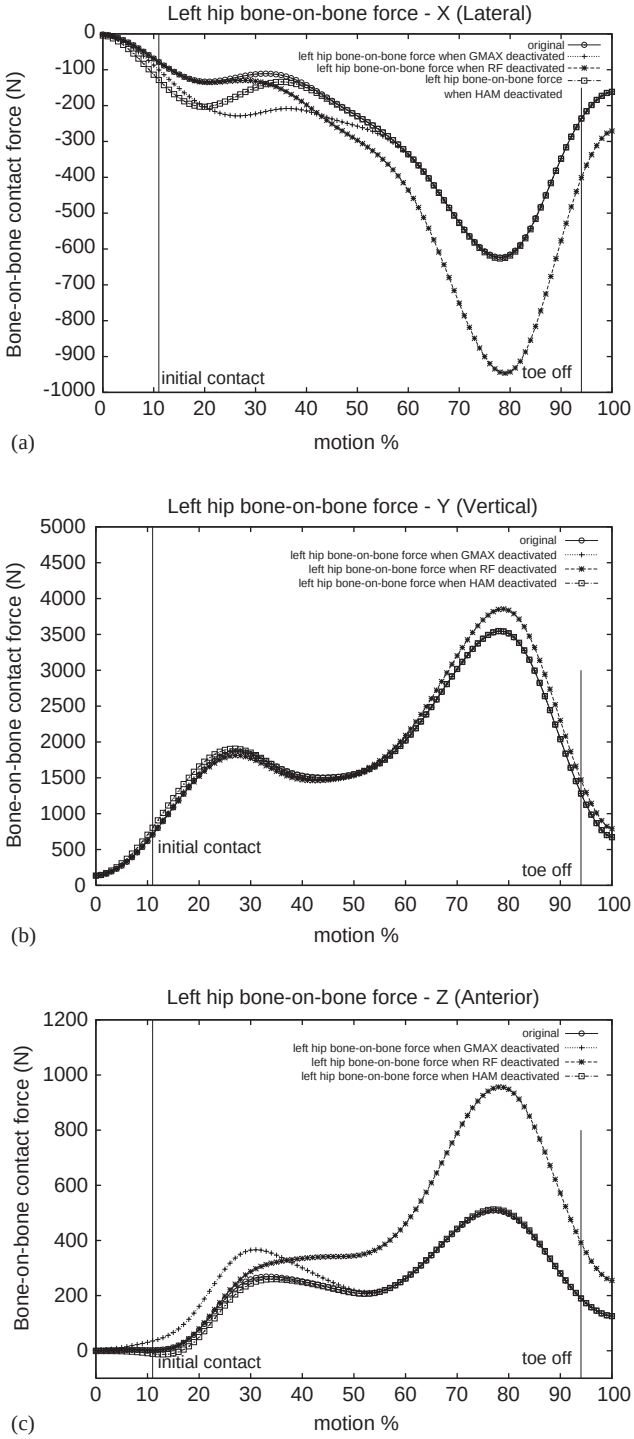

Fig. 5. Bone-on-bone force at the hip along the (a) *X*- (lateral), (b) *Y*- (vertial), and (c) *Z*- (anterior) axis: The original data (shown by ○), the data when the GMAX is deactivated (+), when the RF is deactivated (*), and when the HAM is deactivated (□) are shown.

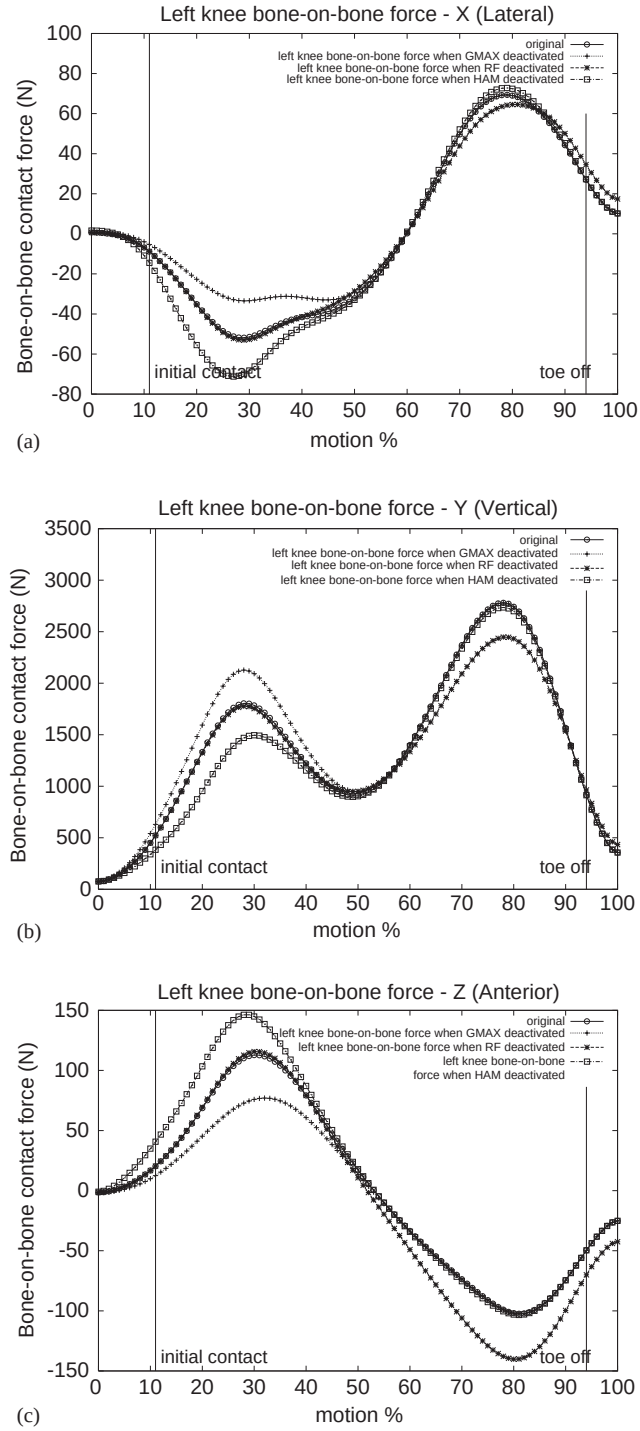

Fig. 6. The bone-on-bone force at the knee along the (a) *X*- (lateral), (b) *Y*- (vertial), and (c) *Z*- (anterior) axis: The original data (shown by ○), the data when the GMAX is deactivated (+), when the RF is deactivated (*), and when the HAM is deactivated (□) are shown.

similar even though the timing of the peak and that of deactivation of the curves often do not match. This might be due to the difference in motions. The patterns are also similar to the muscle force patterns calculated by the static optimization method in studies by Anderson and Pandy (2001) and Crowninshield and Brand (1981).

The activation pattern of the RF calculated in the present study (Fig. 3) might be controversial. The results obtained by using the inverse dynamics method showed

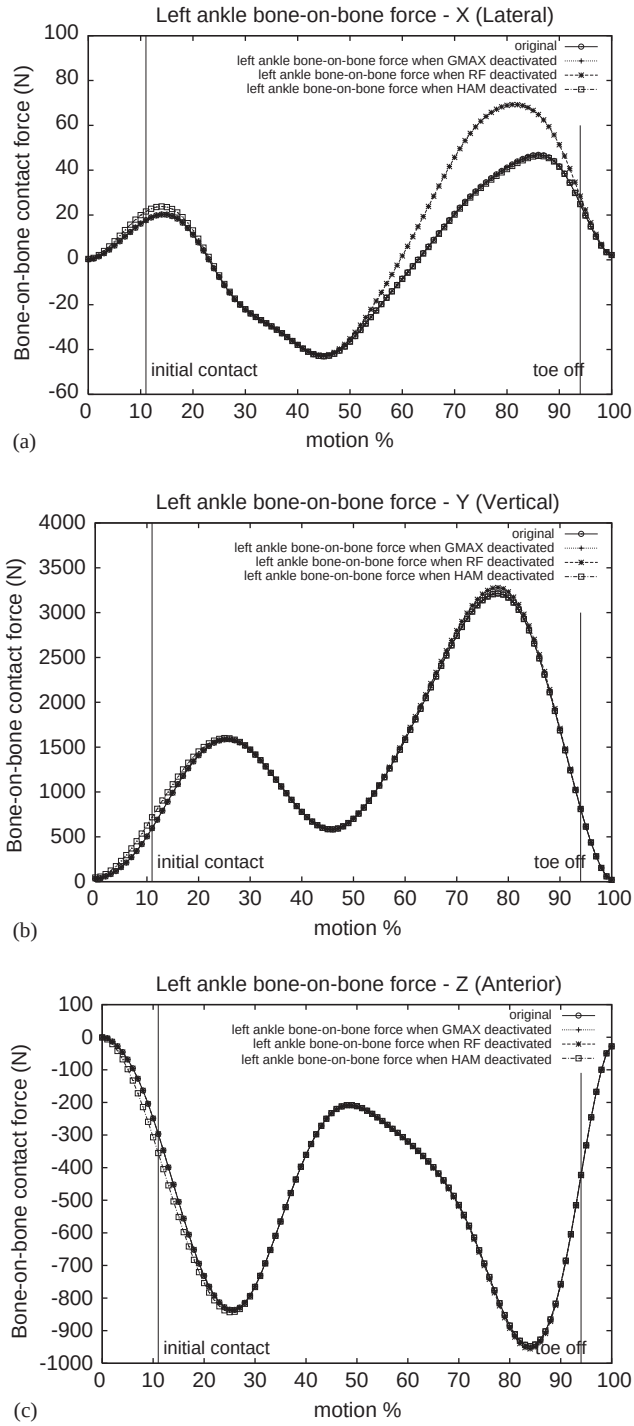

Fig. 7. Bone-on-bone force at the ankle along the (a) *X*- (lateral), (b) *Y*- (vertical), and (c) *Z*- (anterior) axis: The original data (shown by ○) the data when the GMAX is deactivated (+), when the RF is deactivated (*), and when the HAM is deactivated (□) are shown.

that the force exerted by the RF tends to have two peaks, a smaller one in first half and a larger one in the latter half of the support phase (as presented in Anderson and Pandy, 2001). However, the reported EMG patterns of the RF during the support phase are not consistent. For example, Knutson and Soderberg (1995) reported that the RF is active in the first half of the support stage rather during the ending. Therefore, it is difficult to make a conclusion here regarding the validity of the pattern by the RF. For the other muscles, the force patterns and the EMG data patterns obtained in other studies matched well (Fig. 3). Therefore, it seems reasonable to conclude that gait motion data, musculoskeletal model tuned to the body of the patient, and the static optimization method, are sufficiently valid to be used for further experiments in this field of research.

Damage to any muscle of the musculoskeletal system results in the patient either (1) compensating for the damaged muscle by retraining the other muscles to keep the same motion or (2) changing the motion drastically in order to reduce the work by the damaged muscle. The method proposed here assumes that patients do not greatly modify the general patterns of motion (option (2)). Even when patients have to select option 2, it is reasonable to assume that they prefer not to continue performing motions that are substantially different from normal ones.

One of the most distinguished properties of the human body is its plasticity. This means that mechanical properties of the musculoskeletal system can adapt to imposed tasks so that the system can perform the task in a biologically optimal way (Lieber, 1992). In modern civilization, aesthetic and social factors have significant effects on daily life. For example, some healthy "patients" choose to undergo surgical operations of the musculoskeletal system for purely aesthetical reasons (Lemperle and Exner, 1998; Tsai et al., 2001). Therefore, in clinical situations, it is reasonable to assume that patients whose neuromusculoskeletal system has been damaged (e.g., paralysis of some muscles) will choose option (1), i.e., performing the same (or very similar) motion as the one that was employed before the damage occurred (Morita et al., 1998).

Orthopedics considerations will normally result in the patient being asked to select option 1, because a radically different motion involving unusual force that has to be exerted by the ligaments, muscles and bones can cause even greater damage to the body.

In the case of a sports injury, the patient will select option 1, if possible, because the radically different motion resulting from option (2) will degrade performance.

Considering effects those resulting from option (2) requires an optimization technique based on forward dynamics (Yamaguchi and Zajac, 1990; Pandy et al., 1995; Neptune and van den Bogert, 1998; Piazza and Delp, 1996) or a method such as that proposed by Kaplan and Heegaard (2001). However, in applying such methods, it is necessary to prepare a proper objective function that will achieve the motion being

Table 1
Dependence of the muscles on each other for gait

—	GMAX	GMED	GMIN	ILIOP	HAM	ADDL	AMAG	TFL	RF	VAS	GAS	SOL	TIBANT
GMAX	—	36	6	−2	44		4		−7	104			−1
GMED	24	—	110	−22	23	−3	−5	20		28	−3	3	−2
GMIN	5	109	—	51	5	−2	−1	18	40		21	−9	−1
ILIOP	−2	−29	14	—		4		27	159	−17	88	−40	
HAM	35	17	4	21	—		9	1	−11	−34	22	−6	−6
ADDL		−1	−1	34	−2	—	10		6		2		
AMAG	5	−3	−1		13	5	—		−1	50			
TFL		12	9	16				—	17	−4	7	−3	
RF	−3	5	8	66	−3	1		10	—	124	−74	35	
VAS	27	5	−2	−11	−22		7	−3	135	—	−6	58	−1
GAS		−3	8	120	22			8	−49	−7	—	221	−1
SOL		2	−7	−112	−1			−7	45	6	195	—	
TIBANT					13								—

The deactivated muscles are listed horizontally at the top of the table, and the integral of the finite differences in the muscle force are listed vertically. The unit for each value is Newtons (N).

Table 2
Dependence of the joint bone-on-bone force on the muscle force for support phase during the gait

—	GMAX	GMED	GMIN	IIOP	HAM	ADDL	AMAG	TFL	RF	VAS	GAS	SOL	TIBANT
Hip	−1	−42	55	−120	0	0	−2	41	79	147	58	−24	−9
Knee	52	37	35	249	−53	1	14	−17	−41	−84	−273	323	−4
Ankle	0	0	−2	−32	20	0	0	−2	13	1	55	30	20

The deactivated muscles are listed horizontally at the top of the table, and the integral of the norm of the differences in the bone-on-bone force are listed vertically. The unit for each value is Newtons (N).

analyzed. It is known that results obtained by using optimization based on forward dynamics are very sensitive to the objective function. Although an objective function based on minimal energy or maximal height is used for motions such as gait (Anderson and Pandy, 2001) or maximal jumping (Pandy et al., 1990), this cannot be applied to arbitrary motions recorded by a motion capture device. It is still not clear what kind of objective function is appropriate for arbitrary motion. In addition, such methods require a great amount of calculation that will make it even more difficult to determine the effect of a change in arbitrary muscle force on other muscle forces or on joint bone-on-bone forces.

In contrast, the new method proposed here, which is based on inverse dynamics, can be applied to arbitrary motion as long as the static optimization method reproduces the proper muscle activation level. Anderson and Pandy (2001) reported that the results obtained by using inverse dynamics were almost the same as those obtained by forward dynamics in the case of motion such as gait motion. The calculation time for the newly proposed method is short in comparison with the calculation times of other optimization methods based on forward dynamics. Although it seems unrealistic that the human body will generate exactly the same gait motion when some large muscles such as the GMAX, GMED, and SOL are paralyzed, the results of our research suggest that this is possible from the viewpoint of the musculoskeletal system. After an injury or paralysis, the neural patterns transmitted from the central nervous system and the feedback loop of the neurons are same as the signals before the injury despite the fact that the physiological condition of the body has greatly changed. As a result, the motion realized by these signals would surely be different from the original one. In addition, the motion will involve pain in the case of injury, and the human body will therefore intentionally avoid the original motion. However, as the body goes through further training and rehabilitation, the dynamical roles of the deactivated muscle can be compensated by other muscles in most cases. Therefore, it is necessary to know how a motion similar to the original one can be regenerated without using the deactivated muscle. It is possible to determine by using our method which muscles must be trained in order to compensate the role of the deactivated muscles.

To further assure the validity of our method, it is necessary to confirm that the results obtained by using the static optimization method are insensitive to small changes in the movement pattern. We therefore conducted an experiment to determine the sensitivity of

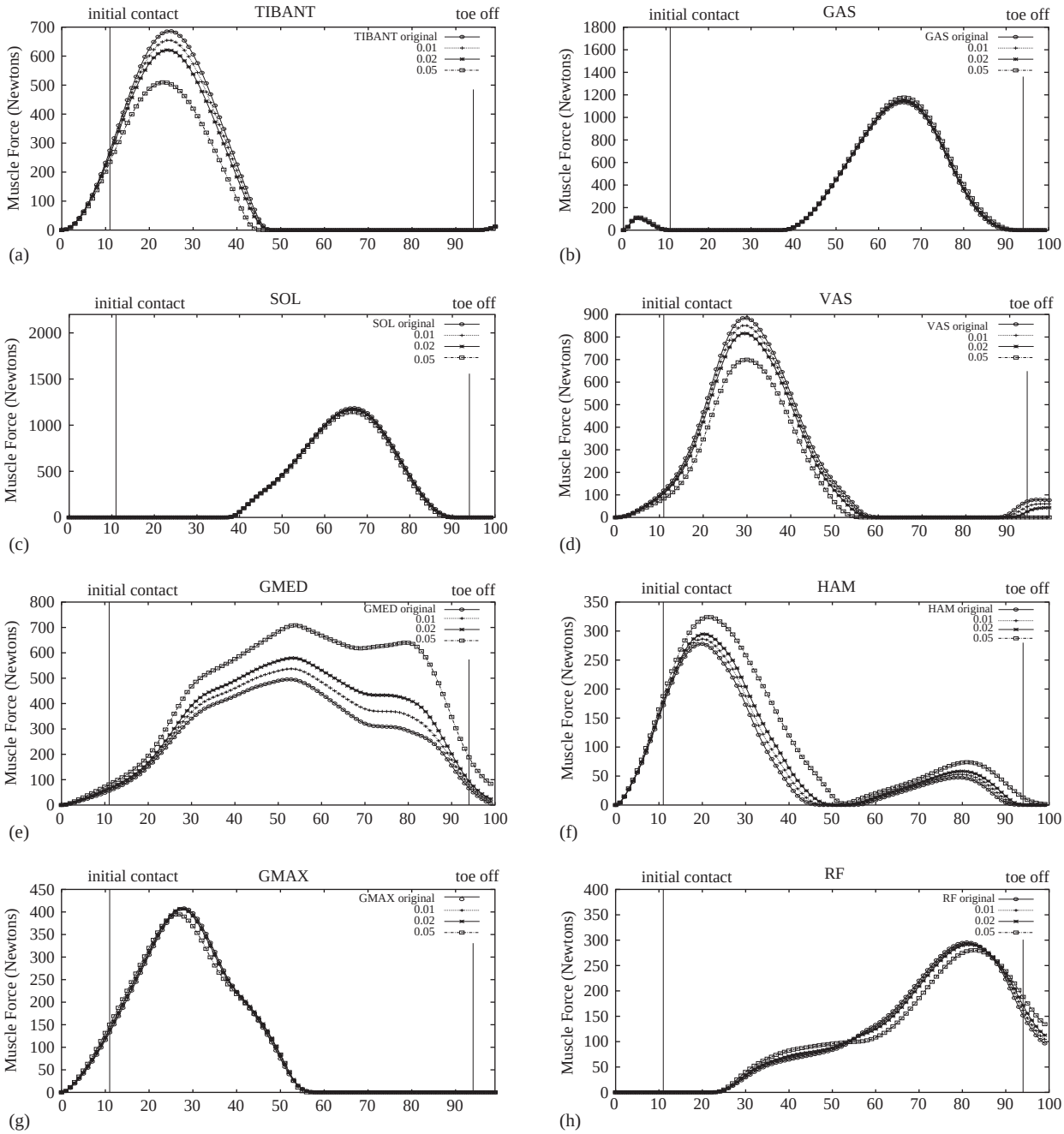

Fig. 8. The muscle force sensitivity to the changes in the movement pattern. Each figure lists the characteristics of (a) TIBANT, (b) GAS, (c) SOL, (d) VAS, (e) GMED, (f) HAM, (g) GMAX, and (h) RF.

results obtained by using the static optimization method to changes in input movement data. Using the same gait motion data as those used in the previous experiment, we parallelly translated the human body model along the X- (lateral), Y- (vertical), and Z- (anterior) axis through the motion while leaving the position of the support foot the same. The translation of the center of pressure of the ground force, $\Delta c'$ was calculated by projecting the translation of the position of the center of gravity, Δc, to the ground (Fig. 9):

$$\Delta \boldsymbol{c}' = \Delta \boldsymbol{c} - (\Delta \boldsymbol{c} \cdot \boldsymbol{e}_y)\boldsymbol{e}_y.$$

By this process, we can obtain a slightly different motion from the original one that satisfies Newton's law of motion. Then the static optimization method was used to calculate the muscle force of the support leg. The muscle force was compared with those calculated from the original data, and the sensitivity of the muscle force

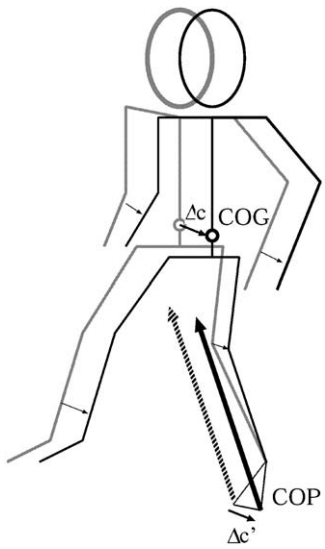

Fig. 9. In order to conduct sensitivity analysis of changes in muscle force data in response to the changes in the movement pattern, the motion data were transformed by moving the COG Δc while keeping the position of the support foot the same. The transfer of COP $\Delta c'$ was calculated by projecting Δc to the ground.

to changes in motion data was estimated. Examples of calculation of the muscle force by setting $\Delta \boldsymbol{c}$ to (0.01, 0, 0), (0.02, 0, 0) and (0.05, 0, 0) are shown in Fig. 8. We also conducted a sensitivity analysis by setting $\Delta \boldsymbol{c}$ to (0, Δy, 0), where $\Delta y = -0.01, -0.02, -0.05$ (only negative values being set to Δy to prevent the distance between the hip joint and the joint angle from being longer than the sum of the lengths of the thigh and shank), and to (0, 0, Δz), where $\Delta z = \pm 0.01, \pm 0.02, \pm 0.05$. A linear regression analysis between Δx and ΔF was carried out, and the average correlation coefficient of the muscle force was found to be larger than 0.99. Therefore, we can assume that the muscle force calculated by the static optimization method will not be drastically affected by small changes in movement data. We therefore conclude that this new method is sufficiently practical for evaluating the effect of a change in muscle force on other parts of the body.

In summary, we have proposed in this paper a method for evaluating the dependence of muscles on each other during arbitrary motion. The method is based on inverse dynamics and optimization. The dependence of each muscle was calculated for a gait motion, and a matrix representing such dependence was established. The use of this matrix makes it possible to determine how the muscles augment each other when any muscle cannot exert its normal force because of injury or paralysis. The influence of deactivated muscle force on joint reaction force was also calculated, and the results showed how deactivation of an arbitrary muscle can result in serious damage to joints.

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