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Functional roles of lower-limb joint moments while walking in water

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Abstract

Objective. To clarify the functional roles of lower-limb joint moments and their contribution to support and propulsion tasks while walking in water compared with that on land.

Design. Sixteen healthy, young subjects walked on land and in water at several different speeds with and without additional loads.

Background. Walking in water is a major rehabilitation therapy for patients with orthopedic disorders. However, the functional role of lower-limb joint moments while walking in water is still unclear.

Methods. Kinematics, electromyographic activities in biceps femoris and gluteus maximums, and ground reaction forces were measured under the following conditions: walking on land and in water at a self-determined pace, slow walking on land, and fast walking in water with or without additional loads (8 kg). The hip, knee, and ankle joint moments were calculated by inverse dynamics.

Results. The contribution of the walking speed increased the hip extension moment, and the additional weight increased the ankle plantar flexion and knee extension moment.

Conclusions. The major functional role was different in each lower-limb joint muscle. That of the muscle group in the ankle is to support the body against gravity, and that of the muscle group involved in hip extension is to contribute to propulsion. In addition, walking in water not only reduced the joint moments but also completely changed the inter-joint coordination.

Relevance

It is of value for clinicians to be aware that the greater viscosity of water produces a greater load on the hip joint when fast walking in water.

Keywords: Walking in water; Functional role of lower-limb joint moments; Synergistic hip extension muscles

1. Introduction

Supporting the body against gravity and generating movements to propel the body forward are considered the two major tasks of the lower limbs for bipedal walking in humans. In general, the coordination of the hip, knee, and ankle joints is required for bipedal walking. From the viewpoint of system control theory, bipedal walking occurs by repeating open- and close-chained-link kinematics, which permits control while walking. In fact, the lower-limb kinematic patterns while walking on land are similar to those of a backward gait (Grasso et al., 1998) and those of walking in water (Miyoshi et al., 2003). The spatiotemporal patterns of the lower-limb kinematics are robust at the stance phase while walking in water, however, the kinetics is quite different

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from what occurs while walking on land (Miyoshi et al., 2003).

It is much more advantageous to walk in water than on land. In water, the force on the joints is easily controlled by changing the submersion level, walking speed, or other conditions. Buoyancy decreases the vertical component of the ground reaction force while walking in water (Nakazawa et al., 1994b). Simultaneously, the viscosity of water requires a greater propulsion force than that required when walking on land (the viscosity of water is at least 800 times that of air). Although it is clear that all movements follow Newton's laws, the functions of lower-limb joint moments while walking in water are quite different from those on land. Clinicians would be better able to determine safe and effective exercise programs if they had knowledge of the differences in the dynamics between walking on land and in water.

There are some general agreements on the interpretation of the function of lower-limb joint moments while walking on land. The function of the knee joint moment is basically to absorb rather than to generate energy and to make the gait smoother rather than to control or propel it (Winter, 1990). In the ankle joint moment, the plantar flexion moment increases and dominates from 10% to approximately 50% of the gait cycle to resist and control the forward rotation of the tibia over the foot (Winter et al., 1995; Czerniecki, 1988; Ericson et al., 1986; Winter, 1991). The hip joint moment acts against gravity to prevent the collapse of the stance limb at the middle and late stance phase (Sadeghi et al., 2001a; Eng and Winter, 1995). However, no experimental data are available concerning the differences between walking in water and on land in terms of the main functional role and interaction of the ankle, knee, and hip joint moments to describe how they support the body against gravity and propel it. Hence, the main objective of this study was to determine the functional roles of the hip, knee, and ankle joint moments while walking in water and to compare them with those while walking on land at different speeds and/or loading conditions. Further, our preliminary data showed that there were opposite kinetic pattern around the hip joint while walking in water compared with those on land (Miyoshi et al., 2003). Thus, another objective was to define the electromyographic (EMG) activities of the hip extensor muscle group while walking in water.

2. Methods

2.1. Subjects

Twelve male and four female healthy subjects (mean (SD) 22.3 (2.7) years, 171.9 (6.2) cm height, 62.8 (9.6) kg weight) participated in this study. They had no past history of orthopedic or neurological ailments, and they had had no recent injury or surgery that could affect their walking and upright standing patterns. All subjects gave their informed consent to the experimental procedures, which were approved by the ethics committee of the National Rehabilitation Center for Persons with Disabilities.

2.2. Experiments

The water temperature was 34 °C, and the water depth, which was at axillae level, corresponded to approximately 20% of the body weight (BW). Before the walking trials, the subjects maintained a standing position for 30 s in water or on land. The vertical component of the ground reaction force (Fz) during quiet standing under water was measured by a waterproof force platform (9253A12, Kistler). From the Fz data, the buoyancy reduced body weight by approximately 80%, since the actual body weight in water while standing was 12.2 (SD 2.3) kg (a body weight reduction of almost 80%). When the walking trials began, the subjects walked in water or on land at a self-determined pace with no loads (control trial) along the walkway. They stepped on a force plate and walked at various speeds with or without weights. Walking trials at various speeds followed. Weights (8 kg) were used for each condition. Walking on land was repeated 15 times for each condition, and walking in water was repeated 10 times for each condition. A total of 100 trials were accepted for the data analysis for each subject. The three components of the ground reaction forces (vertical, anterior–posterior, and medio-lateral) were recorded at the stance phase while walking on land and in water.

2.3. Motion analysis

The walking and the upright standing data were assessed with a two-camera video-based system synchronized with the force plate. Both cameras were placed on the right side of the subject at an average distance of 2.0 m, along an arc of about 45°. This allowed coverage of one stride at a 30 Hz frame rate. Motion analysis software (DIPP-MOTION XD, DITECT Co., Ltd.) was used to reconstruct the marker positions into three-dimensional coordinates. A three-dimensional four-body segment model, consisting of the trunk (trunk + pelvis), thigh, shank, and foot, was defined using five markers with a 3 cm-square seal (0.1 mm thickness) at the following landmarks (Miyoshi et al., 2003): the midpoint on the iliac crest between the anterior and posterior superior iliac spine, the greater trochanter, the lateral femoral condyle, the lateral malleolus, and the fifth metatarso-phalangeal joint. All markers were waterproof, and they were placed over the right side of the leg and trunk. For each joint, the center of rotation was estimated at the geometric center of the joint. A fourth-order zero-lag Butterworth filter was applied to reduce the noise in the three-dimensional coordinates (cut-off frequency: 3 Hz).

2.4. EMG recording

We measured the EMG activities while eight subjects walked in water. A pair of surface electrodes (Ag–AgCl, 6 mm in diameter, 3 cm in interelectrode distance) was placed on the gluteus maximums (GLM) and the long head of biceps femoris (BF) muscles of the right leg, and EMG signals were amplified using a bioamplifier (MEG 6108, NIHON KODEN Co., Ltd.) with a band pass filter from 30 Hz to 5 kHz. For the site of each pair of electrodes, one was placed on each muscle berry and the other was above 3 cm along to each muscle fiber. An adhesive microfilm (0.1 mm thickness) was used for the waterproof treatment of the electrodes. All analog signals (EMGs and ground reaction forces) were digitized at a sampling rate of 1 kHz (WE7251; Yokogawa Electric Co., Ltd.) for off-line analysis.

2.5. Data analysis

Body segment parameters, as well as kinematic and force plate data, were used in an inverse dynamic approach (the methods of which were demonstrated by Winter (1990)) to calculate the net muscle moment at the hip, knee, and ankle joints in the sagittal plane throughout the stance phase. The joint moments were expressed where the flexion (hip), extension (knee), and dorsiflexion (ankle) moments were considered positive. For averaging purposes, the joint moment data were normalized to 100% of the stance phase in each walking condition and with respect to each subject's body weight and the length of the lower-limb segment. The lower-limb length was calculated from the vertical coordinate of markers between the great trochantar and the fifth metatarso-phalangeal joint (86.7 (SD 4.1) cm).

We constructed the three-dimensional (3D) plot of the joint moment (z-axis) as a function of the walking speed and the ground reaction force. The walking speed was demonstrated as a propulsion task, and the ground reaction force, as a task in response to gravity. To develop the 3D plot, the stance time (ST) and the impulse (the area under the Fz curve) were measured during the stance phase in each walking condition; the Fz (mFz) was averaged and calculated from the impulse from the ST. The averaged joint moment of each joint at the stance phase while walking in water was also calculated. With regard to the walking speed, the inverse of the ST value was presented as a speed index (SI). All three of these parameters were presented as relative values to those under the control trial. Each of the joint moments can be expressed as a single plane on the 3D plot. To analyses this 3D plot, two slopes of each single plane were used to quantify the contributions of the walking speed and the ground reaction force to the joint moments.

At first, all of the EMG time series were full-wave rectified. We tried to predict the time series of each muscle torque by applying a 6-D polynomial equation to each EMG activities, since the 6-D polynomial regression was above 0.7 of the regression coefficient firstly in all EMG time series. Then, the predicted hip joint moment was calculated in the following manner: {(predicted hip joint moment) = (predicted BF torque time series)$*s$ + (predicted GLM torque time series)$*t$ + (predicted the other agonisic muscles)$*u$ + (predicted antagonistic muscles)$*v$}. To the best of our knowledge, the distribution gain of the resultant net joint moment to each muscle torque is still unknown. Therefore, in this simulation, we applied the gain $\{(s,t,u,v) = (1,1,0,0)\}$ in above.

2.6. Statistics

A student t-test was used to determine the contribution of the load or the walking speed for each joint moment. Differences were considered statistically significant at $P < 0.05$.

3. Results

Although each subject determined his walking speed, the speed calculated from the displacement of the marker on the greater trochanter in each condition (mean (SD)) was 1.03 (0.11) m/s for a comfortable speed and 0.54 (0.13) m/s for slow walking on land; 0.55 (0.09) m/s for comfortable walking in water and 0.93 (0.15) m/s for fast walking in water.

The typical joint moments are demonstrated in Fig. 1. The mean (SD) was calculated at the ankle (top), knee (middle), and hip (bottom) in the sagittal plane at the stance phase while walking on land and in water in one subject. The hip joint moment was shown to be an extension moment throughout most of the stance phase while walking in water, whereas, on land, it changed remarkably from an extension to a flexion direction. Without reference to the additional loads, the mean (SD) hip extension moments while walking in water increased as the walking speed increased (comfortable: 0.051 (0.004); fast: 0.075 (0.010) without additional loads; comfortable: 0.053 (0.005); fast: 0.076 (0.016) with additional loads, $P < 0.05$). There were no significant changes in the hip joint moments while walking on land and increasing speed; however, the hip flexion moments were significantly increased when weight was attached. The knee joint moment had two extension peaks as the subjects walked on land, whereas, in water, only one extension

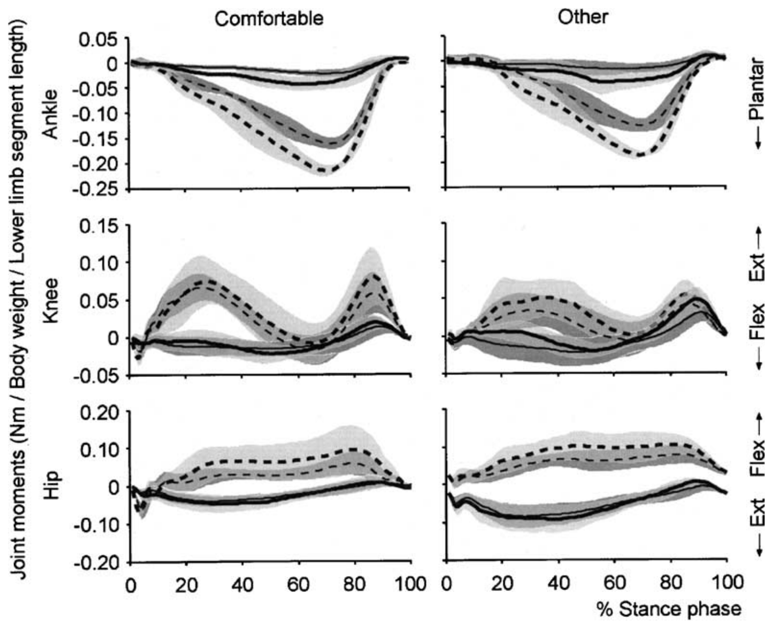

Fig. 1. Averaged hip, knee, and ankle joint moments in the sagittal plane at the stance phase while walking on land (dashed line) and in water (solid line) for one subject. Left panels: all of these lines indicate a comfortable walking speed; thin lines denote no additional loads; and thick lines denote additional loads attached. Right panels: dashed lines denote walking on land at a slow speed; solid lines denote walking in water at a fast speed, and the explanation for each line is the same as that in the left panel.

peak appeared at the late stance, and the other was a flexion moment. The early extension moment in the knee joint appeared when the walking speed was increased while walking in water, and these two extension peaks increased slightly but without significance when weight was attached. The ankle joint moment curve while walking in water was considerably reduced but in the same (plantar flexion) direction. These plantar flexion moments while walking on land and in water apparently increased when the weight was attached and/or when the walking speed was increased.

A typical example of the relationship of each joint moment, mFz, and walking speed (SI) is shown in Fig. 2. The contribution of the load and the walking speed to each joint moment is denoted by two gains of the single plane on the 3D plot as a function of both the mFz ratio and the SI. The knee and the ankle joint moments while walking in water and on land had similar tendencies, since the gain for the relation of the mFz ratio was 0.072 while walking on land and 0.402 while walking in water for the knee and 1.094 and 0.648 for the ankle, while, for the relation of the SI, it was 0.718 and 0.584 for the knee and −0.546 and −0.509 for the ankle, respectively. For the hip joint moment, however, the gain for the relation of the mFz ratio was 0.645, and that for the relation of the SI was −0.459 while walking on land; on the other hand, while walking in water, the gain for the relation of the SI was 0.902. These tendencies, which were observed in all subjects, indicated that the weight and walking speed affected the knee and the ankle joint moment in a similar manner while walking on land and in water; however, the weight and walking speed affected only the hip joint moment while walking in water. These results show that the function of the hip and knee joint moment was clearly different while walking in water than while walking on land.

Fig. 3 demonstrates typical EMG activities in one subject at the stance phase while walking in water. The waveform of the GLM EMG activity increased slightly, but its patterns did not change as the walking speed increased. In the case of the BF, although the GLM and BF are both a part of the synergistic hip extensor muscle, the amplitude and the time-series at the stance phase while walking in water, especially from the early to the mid-stance phase, were clearly different when the walking speed increased. We predicted each muscle torque (middle column) by applying a 6-D polynomial equation to each EMG time series. The predicted net hip joint extension moments (right upper column) were also simply calculated by the summation of each predicted muscle torque. The result clearly shows that these predicted hip joint extension moments were well matched to the net hip joint extension moment time series throughout the stance phase while walking in water with an increase in the walking speed.

4. Discussion

The main objectives of this study were to identify the main function of the hip, knee, and ankle joint moments and their contribution to the function of support and

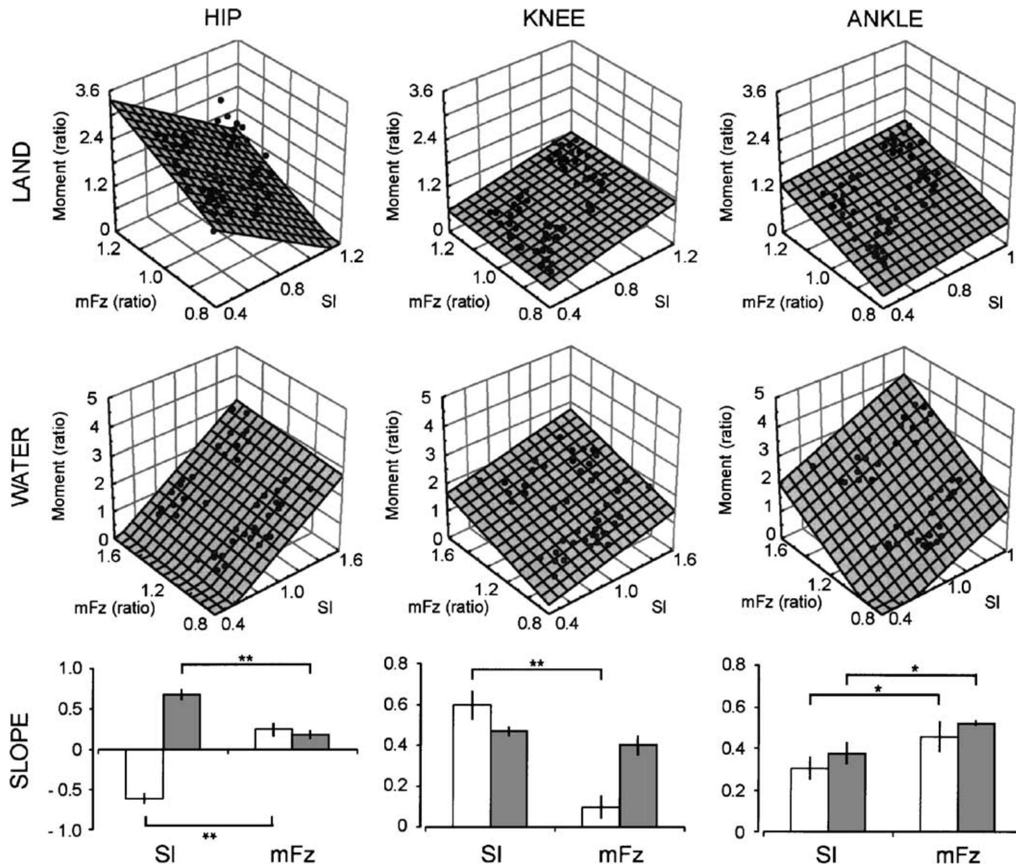

Fig. 2. Typical 3D plots between the joint moment ratio and the parameters of the mFz ratio and SI while walking on land (top panels) and those in water (middle panels), using the same subject as that in Fig. 1. Top: while walking on land; Middle: while walking in water. The contribution of either mFz or SI to each joint moment is demonstrated in the bottom panels. The white bars indicate walking on land, and the gray bars indicate walking in water. While walking in water, all of the joint moments increased as the walking speed increased, whereas the hip joint moment was independent of any additional loads. In the hip joint moment while walking in water, the function was apparently different from that on land. This situation is shown in the bottom panels. ($^{*}P < 0.05$, $^{**}P < 0.01$).

propulsion at the stance phase and to compare the results from walking in water to those with walking on land. Consequently, the main results of this study are as follows: (1) The ankle plantar flexion moment was significantly decreased while walking in water. (2) The knee extension moment waveform while walking in water had only one extension peak at the late stance in spite of the two extension peaks that appeared while walking on land. (3) The hip extension moment appeared almost throughout the stance phase while walking in water. (4) The hip extension moment increased with an increase in the walking speed while walking in water. (5) The BF EMG activities while walking in water increased, and the waveforms clearly varied with an increase in the walking speed, whereas the GLM EMG activities increased slightly, but the waveforms were independent of the walking speed.

4.1. Functional role of the ankle joint moment

It is well known that the function of the ankle joint moment at the stance phase while walking can be described as follows: (1) Dorsiflexion moments, which usually occur in the first 10% of the gait cycle, are for the control of lowering the foot (Winter et al., 1995). (2) The plantar flexion moment increases and dominates from 10% to approximately 50% of the gait cycle to resist and control the forward rotation of the tibia over the foot (Winter et al., 1995; Czerniecki, 1988; Ericson et al., 1986; Winter, 1991). There is some controversy regarding the function of the ankle plantar flexion moment while walking. The focus of a major controversy regarding the function of ankle plantar flexion moments is at the late stance phase. These controversy includes the following issues: (1) The main source of energy propels the trunk upward and forward after heel-off and in many cases until toe-off (Winter, 1983). (2) The function is to restrain, not accelerate, the trunk over the ankle in walking (Sutherland et al., 1980; Sadeghi et al., 2001b). (3) It acts as an accelerator, which facilitates the movement of the leg into the swing phase (Dillingham et al., 1992; Meinders et al., 1998).

From the viewpoint of the system theory, the stability of the stance phase may operate in the same manner as

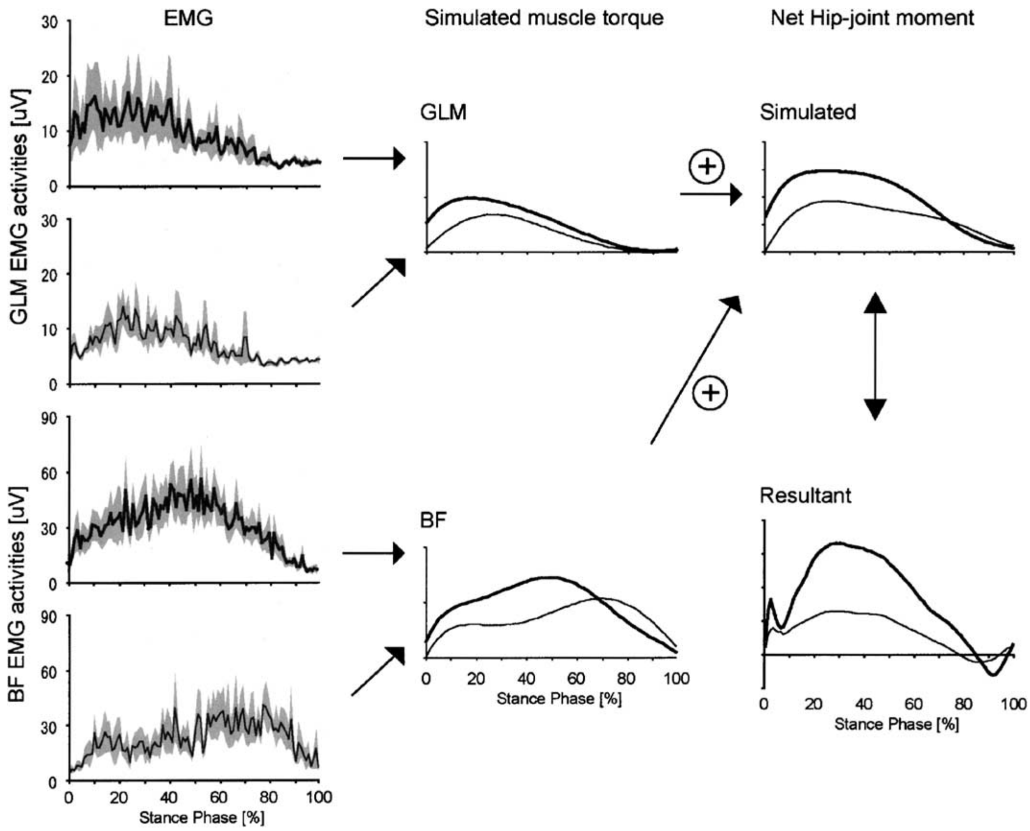

Fig. 3. Left panels: Typical EMG activities at the stance phase while walking in water, using the same subject that is shown in Fig. 1. The top two figures indicate the GLM EMG activities, and the bottom two, the BF EMG activities. Center panels: Simulated muscle torque in GLM (upper) and BF (lower). Right panels: Simulated net hip joint moments (upper) and resultant net hip joint moments (lower). Thin lines denote a comfortable walking speed, and thick lines, the conditions for fast walking in water. The predicted hip joint extension moments were close to the net hip joint extension moment time series throughout the stance phase while walking in water with an increase in the walking speed.

the stability of an inverted pendulum. The ankle plantar flexion moment increases, since the postural instability increases accompanied by an increase in the load. On the other hand, the ankle plantar flexion moment increases accompanied by an increase in the walking speed, since it propels the body forward. This situation is demonstrated in Fig. 2, which shows that the ankle plantar flexion moment had high sensitivity to the weight and less sensitivity to the walking speed. Therefore, these results suggest that the main function of the ankle joint moment is to maintain stability against gravity. Furthermore, they strengthen the idea (Sutherland et al., 1980; Sadeghi et al., 2001b) that the function of the ankle plantar flexion moment is not to accelerate the body.

4.2. Functional role of the knee joint moment

There are two extension peak moments, called double knee action, at the knee joint at the stance phase while walking on land (Kawamura et al., 1999). The first extension peak moment, especially, was conceived to be useful for absorbing the impact force acting on the early stance phase while walking on land (Kawamura et al., 1999). In the present study, the impact force while walking in water was drastically reduced due to buoyancy. However, the impact force in both conditions increased with the weight load and/or increase in the walking speed. The early burst of the knee extension moments were drastically decreased while walking in water compared with those on land (Fig. 1). While walking in water at fast speed, the later knee extension moment slightly increased but had no significance. Therefore, the results also suggest that the function of the knee joint moment is to absorb the impact force during gait.

4.3. Functional role of the hip joint moment

While walking on land, the hip extension moment, which occurs shortly after heel-strike, has been associated with control of the forward acceleration of the trunk (Winter, 1983). At the middle and late stance phase, the hip joint moment acts against gravity to prevent the collapse of the stance limb (Sadeghi et al., 2001a; Eng and Winter, 1995), whereas it should also be considered as a source of propulsion (Sadeghi et al., 1997; Winter and Eng, 1995).

Walking in water simultaneously decreased the load and increased the force to move the body forward, due to the water's buoyancy and resistance, respectively. Stronger propulsion is needed to propel the body forward against the water resistance while walking in water. According to Nakazawa et al. (1994a), biceps femoris muscle activities increased as the walking speed increased while walking in water. Furthermore, these researchers suggested that the hip extension moments while walking in water increased in order to propel the body forward. The results obtained here show that the hip joint moment is an extension moment throughout the stance phase and the peak hip extension moment increases as the walking speed increases while walking in water. This situation was shown clearly in Fig. 2, which shows that the hip joint moment while walking in water is counter to what occurs while walking on land. The hip joint moment while walking in water was more sensitive to the walking speed and less sensitive to the weight loads. The results suggest that the main function of the hip extensor moment throughout the stance phase is to propel the body forward against the water resistance. The findings strengthen the idea that the hip extensors should be considered as a source of propulsion (Sadeghi et al., 1997; Winter and Eng, 1995).

4.4. Neural mechanisms

Fig. 3 shows that these predicted hip joint extension moments are convenient for explaining the net hip joint extension moments and their enhanced peaks with the increase in walking speed. At the stance phase while walking in water, the function of both the BF and GLM muscle should be in common framework generating the propulsive force around the hip joint, van Ingen Schenau et al. (1992) and Jacobs and van Ingen Schenau (1992) have suggested that the bi-articular muscle plays a role in control of the direction of the force. Therefore, our results might indicate that the bi-articular muscle BF generates larger hip extension torque, whereas the mono-articular muscle GLM tunes the hip extension torque in exactly. In our simulation, we demonstrated the GLM and long head of BF EMG activities to clarify the functional role of the hip joint moment while walking in water, whereas we did not measure either the short head of BF, semitendinosus muscle or antagonistic muscles (hip flexor muscles). Future studies will clarify whether there are such differences in the neuromuscular features of synergistic hip extension muscles and their relation to motor functions.

4.5. Clinical relevance

We described how our results could be applied to walking in water for rehabilitation. Patients with knee osteoarthritis have greater reduced knee extensor moments than normal subjects, which suggests that patients compensate to reduce the load on the knee (Sadeghi et al., 1997). Our findings showed that walking in water could be a useful rehabilitative exercise for reducing the joint load since the knee joint moment was shown to reduce the load. Namely, exercises involving walking in water could be designed to incorporate large-muscle activities, especially those of lower-limb extensor muscles and the minimization of joint loading. However, when a subject increases his walking speed in water, careful attention must be given to avoid any sudden enhancement of the hip joint moment acting in the late stance phase.

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