Investigation of the functional stability limits while squatting

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Abstract
This study investigates the functional stability limits (FSLs) in the squatting positions. Eleven male participants leaned and moved their pelvis horizontally in the clockwise and counter-clockwise directions while squatting at 11 depth levels. The depth was controlled by changing the hip height from 100% to 0% of the upright position. The FSLs and the center of pressure excursion lengths were calculated from the force-plate data, and the musculoskeletal loads on the lower limbs were estimated from the joint torques and surface electromyograms. As the hip height reduced, the area of the FSLs narrowed by up to 20% of the base of support (BOS) area at the deepest squatting position. The narrowing was affected by the decreasing FSLs in the forward direction, which also decreased by up to 20% of BOS. These quantitative data accurately evaluate the postural stability, suggesting a considerable fall risk during tasks requiring the squatting position.

KEYWORDS
center of pressure, functional stability limits, musculoskeletal loads, postural stability, squatting posture

INTRODUCTION

Loss of balance during work is a serious problem, risking slips, trips, and falls (STFs) and other workplace accidents (Hsiao & Simeonov, 2001). According to the statistical reports of occupational accidents, STFs contribute 26% of the workplace injuries in the United States (U.S. Bureau of Labor Statistics, 2018) and 40% of that in Japan (Japan Industrial Safety & Health Association, 2018). STFs often result from loss of balance due to awkward postures such as squatting positions (DiDomenico, McGorry, & Banks, 2016).

Squatting postures are involved in many physical tasks on worksites and require workers to flex knees, hips, and torsos, and dorsiflex ankles (Chung, Lee, & Kee, 2003). These postures place considerable musculoskeletal stress on the workers’ body because the joint torques and muscle exertions of the lower limbs are extremely greater than those in the upright postures (Dahlkvist, Mayo, & Seedhorn, 1982; Jensen, 2008). Moreover, since the flexibilities in motions of lower-limb joints are restricted by the musculoskeletal stress (Kasuyama, Sakamoto, & Nakazawa, 2009), the maintenance of body balance is much more difficult than while standing upright (Sriwarno, Shimomura, Iwanaga, & Katsuura, 2008). Therefore, to estimate the risk of falls for workers while squatting, the postural stability has been evaluated using software such as the digital human simulation systems (Kawano, Onosato, & Iwata, 2003; Kawano, Ueda, Fukui, & Sugimura, 2009).

Postural stability has been studied in the biomechanics and robotics fields and is often defined by the mechanical constraints of bipedal standing for both humans and the biped-robots (Goswami, 1999; Holbein & Redfern, 1997; Winter, 1987). The center of pressure (COP) of the foot reaction forces can move only within the area outlined by the feet in contact with the supporting surface and must act at the almost same horizontal location as the body’s center of mass (COM) to maintain equilibrium (Winter, 2009).
defined as the functional stability region (FSR; Holbein & Chaffin, 1997; the BOS within which individuals are willing to extend their COP is the voluntary movable range of the COP within the BOS. The portion of the BOS within which individuals are willing to extend their COP is defined as the functional stability region (FSR; Holbein & Chaffin, 1997; Holbein & Redfern, 1997) or the functional stability boundary (FSB; Bagchee, Bhattacharya, Succop, & Emerich, 1998). The functional limit of the BOS (FBOS), which is defined as the effectively utilized area for the COP movement, is a concept that is similar to FSR (Fujimoto, Hsu, Woollacott, & Chou, 2013; King, Judge, & Wolfson, 1994). The reduced FSR or FBOS limits an individual’s ability to maintain balance because the COP–COM distance is proportional to the COM acceleration (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). The functional stability limits (FSLs) are defined as the distance that an individual will displace their COP from the center of the BOS to the FBOS in a particular direction (Holbein, McDermott, Shaw, & Demchak, 2007). FSLs and FSR are smaller than the theoretical maximum because they are limited by muscle strength, internal postural control, external load control, and other factors (Holbein & Chaffin, 1997). Moreover, the effects of various factors on FSLs and FSRs have been revealed; among these factors are postural muscle strength, reaction time, feet placement, ability to coordinate body-segment movements, somatotopic differences, age, psychological factors, and load-carrying (Bagchee et al., 1998; Fujimoto, Bair, & Rogers, 2015; Holbein & Redfern, 1997; Qu & Nussbaum, 2009).

FSLs have been used in biomechanical models to predict the horizontal pushing/pulling forces and working postures or movements and when the compensatory stepping is required (Holbein et al., 2007). Few studies, however, have investigated the FSLs while squatting. Although squatting postures are expected to reduce the FSLs, the postural balance of squatting postures is poorly understood. The risk of falls, which is often calculated as the ratio of actual COM or COP movement to FSLs, may be underestimated relative to the actual state. Therefore, to investigate the effect of squatting on the postural balance, this research measures the FSLs of squatting postures at multi-level hip heights (HHs). Furthermore, the musculoskeletal loads on lower limbs are then evaluated to examine the biomechanical constraints while squatting.

2 METHODS

2.1 Participants

Eleven men aged 22–27 years participated in the study. The sample size was determined by a statistical power calculation based on pilot study data (Lee & Lee, 2003). All participants were in good health with no medical history of musculoskeletal injuries within the previous 12 months. The means and standard deviations of the age, height, and weight variables were 22.8 ± 0.9 years, 171 ± 4.0 cm, and 65.0 ± 5.3 kg, respectively. When standing upright, the vertical height from the floor to the greater trochanter was 88.6 ± 4.1 cm. All participants wore the same model of safety shoes (MZ010; Midori Anzen Co. Ltd., Japan). The average foot length of the participants (International Organization for Standardization, 2017) was 260 ± 0.7 cm, and the shoe size ranged from 25.0 to 27.0 cm. After an oral overview of the experiment, each participant gave their informed consent to participate. This experiment was approved by the ethics committee of Tokyo Metropolitan University, Hino Campus (approval number: 238).

2.2 Experimental apparatus and procedure

While FSLs or the limits of stability (LOS) have been measured by various inspection methods, inconsistencies in techniques may have limited the accuracy and reproducibility of the estimates due to leaning speed, allowing loss of foot contact or constructing ellipses from limited data samples (Forth, Fiedler, & Paloski, 2011). Therefore, this study adopted the verified functional limits testing procedure using a controlled low-speed voluntary leaning protocol that requires feet to remain in continuous contact with the ground (Forth et al., 2011; Juras, Slomka, Fredyk, Sobota, & Back, 2008). The participants of this study were required to lean forward as far as possible while maintaining heel contact with a force plate (9286AA; Kistler Instrument Corp., Amherst, NY). After reaching this forward stability limit, they moved their hip horizontally while maintaining a squatting posture for 20 s. Figure 1 shows the flow of their movements. Within the first 5 s, the participants squatted and adjusted their knee height until the hip marker overlapped the level line of the feedback system. They then leaned and moved their pelvis, drawing an imaginary circle in the horizontal plane as widely as possible without changing the HH. Next, they moved in the clockwise direction for 10 s, followed by the counter-clockwise direction for 10 s. Before the experiments, the participants had determined their comfortable foot position for squatting on a force plate. The outline of the standing position was marked on the plate by 10-mm-wide masking tape, and the trajectory was electronically recorded by pressing the tape with a finger. Using the recorded coordinates, the inner area was calculated as the BOS by summing the infinitesimal triangular areas formed by the two neighboring extracted points and the central coordinates. In all trials, participants were asked to maintain the same foot position.

The HH was defined as the vertical height from the floor to the greater trochanter during upright standing. A reflective marker (14-mm diameter) was attached on the left greater trochanter of each participant, and its location was captured from the left side by a video camera. The captured vision was recorded on a computer and projected onto a liquid crystal display placed in front of the participant (Figure 1). Using a video-editing program, a colored line (approximately 15-cm wide) was superimposed on the vision as a reference line of the HH, providing visual feedback to the participant. The height of the camera was adjusted such that the marker on the greater trochanter was always captured directly from the side.
In this study, the HH relative to the body height was controlled as an experimental factor. The vertical heights of the greater trochanter in the upright position and in the squatting position with maximum knee flexion were defined as 100%HH and 0%HH, respectively (Figure 2). The experimental conditions were 11 levels of HH (0%, 10%, 20%, ..., 100%HH). The experimental conditions were completely randomized, and measurements were duplicated under each condition.

2.3 | Measurements and analysis

2.3.1 | Center of pressure and functional stability limits

While the participants squatted and leaned, the force components \( F_x, F_y, \) and \( F_z \) and moment components \( M_x, M_y, \) and \( M_z \) were sampled at 100 Hz by the single force-plate system with an analog-to-digital data converter (PH-703; DKH Co. Ltd, Japan) and analysis software (TRIAS2; DKH Co. Ltd). The signals were low-pass filtered by the moving-average method (1-s average block). As the fast Fourier transformed signals peaked near 0.1 Hz, the cutoff frequency was set to 0.443 Hz. The position-time trajectory of the COP in the horizontal plane was determined through standard transformations (Winter, 2009).

Figure 3 illustrates the data-processing steps. In all ranges of azimuthal angles (each covering \( 6^\circ \)), the farthest COP positions were extracted based on the average point of the COP perturbation. The 60 extracted COP points were connected with smoothed lines, and the perimeter of the resulting closed curve was defined as the FSLs. The inner area of the FSLs was then calculated along with that of the BOS. Moreover, to determine whether the FSL depends on the moving direction, the maximum proximity of FSLs to BOS (MP) were calculated in the forward, backward, right, and left directions (each with an azimuthal range of \( 60^\circ \)) while referring to the relevant research (Bagchee et al., 1998). The MP was defined as follows:

\[
MP = \frac{|\vec{a}|}{|\vec{b}|}
\]

Here \( \vec{a} \) and \( \vec{b} \) represents the position vector of FSLs and BOS, respectively, from the average point of the COP perturbation for the case that the distance between FSLs and BOS (\( = \vec{b} - \vec{a} \)) has a minimum value.

2.3.2 | Lower-limb loads

Muscles transmit forces and create torque through the insertion into the skeletal structure around the joints. The torque is a function of the moment arm of the insertion and the angle at which the muscle is applying the force (Redfern, 1992). Therefore, it is common to see electromyogram (EMG) signals related to the generated torque instead of the internal muscle force. Therefore, this study measured the joint torque and muscle activities during the squatting tasks.

Before calculating the joint angles, the entire body posture was captured by a motion capture system (Perception Neuron, Noitom Ltd., China) using a software (Axis Neuron Ver. 3.8.42.8303, Noitom Ltd., China). The participant wore gloves and body straps with mounted 18 inertial measurement units (IMU). The position and tilt angle of IMU were recorded at 120 fps and converted into the body posture angle as biovision hierarchy (BVH) format files, including the hierarchy of body skeleton and the rotational angles. We configured the whole-body posture based on the BVH format data and estimated the joint torques at the knee and ankle joints using the three-dimensional biomechanical analysis method (Chaffin, 1997; Winter, 2009). The estimated joint torques were then normalized by the maximum voluntary torque on each joint (Chaffin, Andersson, & Martin, 2006).

The resulting values are called the joint torque ratios. The physical loads on the knee and ankle joints during the rotational motion were then determined from the maximum and 1-s-averaged joint torque ratios. The maximum joint torque ratios in the forward, backward, right, and left directions were calculated, as described for the FSLs above.

The activities of six muscles (R and L rectus femoris, R and L gastrocnemius, and R and L tibialis anterior) were recorded by...
surface electromyogram (sEMG) amplifiers (SX230, Biometrics Ltd., UK) placed on the left and right sides of the body. The attachment positions of sEMG amplifiers were determined based on a reference (Criswell, 2008). Before the trials, the sEMG signals during maximum voluntary contraction (MVC) of each muscle were recorded in a manual muscle test (Hislop, Avers, & Brown, 2013). The raw signals were sampled at 1000 Hz and lowpass filtered through a Butterworth filter with a 2-Hz cutoff frequency. Finally, to eliminate the individual differences among the participants, the raw signals were converted to relative MVC values (%MVC).

2.4 | Statistical analysis

The effects of the experimental factors on the measured indices were compared by analysis of variance (ANOVA) with a two-way factorial design (viz., the hip-height conditions and participants) and a post hoc Tukey's test. Sphericity was checked by the Mauchly sphericity test. The statistical significance level of all tests was set to 5%. Data analyses were carried out using BellCurve for Excel version 3.10 (Social Survey Research Information Co., Ltd., Japan).

3 | RESULTS

3.1 | Functional stability limits

The COP trajectories at HHs of 100%, 60%, 30%, and 0% are shown in Figure 4. These trajectories are the average COP coordinates of all participants in each 6° range of azimuthal angles. According to these results, squatting narrowed the area of the FSLs as the HH lowered. The reduction was notable in the forward, left, and right directions, but was slightly in the backward direction (near the line connecting the lateral malleoli).

ANOVA indicated that the FSL area ratios have the main effect of HH. The bar chart in Figure 5 shows the averages and standard deviations of the FSL area ratios under each hip-height condition. The ratios were calculated as the inner area of the COP trajectories divided by the BOS area of each participant. The FSL area was 34.6 ± 4.2% of the BOS in the upright position (100% HH), decreasing to 18.8 ± 8.0% of the BOS at 0% HH. The line graphs superimposed on the bar chart in Figure 5 shows the MP of FSLs to BOS in the four directions. The FSLs ranged from 20% to 40% of the BOS in the forward direction, and from 40% to 70% in the right, left, and backward directions. Concisely, the FSLs were shorter in the forward direction than in the other directions at all HHs.

3.2 | Joint torque ratio

Figure 6 plots the joint torque ratios at the different HHs. The average and maximum knee-torque ratios at each HH are shown in Figure 6a. Both values increased with decreasing HH. Under the 0% HH condition, the maximum and average knee-torque ratios exceeded 50% and 30%, respectively. On the other hand, the maximum ankle-joint torque ratios were below 30% under all HH conditions and did not significantly depend on HH (Figure 6b).
The mean muscular activities on the left and right legs during the LOS tests are shown in Figure 7a–c. Under the 100% hip-height condition, the activities of three muscles were approximately 10% MVC. The muscular activity of the rectus femoris peaked at over 40% MVC at HHs of 60–40%, and decreased at lower HHs (30–0%). On the other hand, the activities of the gastrocnemius muscles were approximately 10% MVC, with no significant variations among the hip-height conditions. Meanwhile, the sEMG values of the tibialis anterior muscles increased with decreasing HH, reaching above 30% MVC at 0% HH.

4 | DISCUSSION

4.1 | Influence of hip height on functional stability limits

This study examined the FSLs of the squatting posture, and calculated the FSL areas, torque ratios of the lower-limb joints, and muscular activities at different HHs. The mean FSL area was maximized at 35% of the BOS in the upright position (100% HH). When the participants leaned and moved their hips in a circular motion, the COP shifted by up to 60% of the BOS boundary in the left and right directions. This trend concurs with the literature (Holbein & Redfern, 1997), in which the center of gravity shifted by 60.3 ± 9.2% to the right and 58.6 ± 11.5% to the left as participants performed similar motions with no load.

The FSL area decreased with lowering of the HH, being minimized at 19% at the deepest squat position (0%HH). As the participants squatted lower, the MP of the COP decreased in the front, left, and right directions, but did not significantly change in the backward direction. Therefore, the decreased FSLs were largely attributable to the reduced moving range of the COP in the forward and sideways directions. These results confirm that the FSLs are affected by squatting postures, and their area reduces to approximately 19% of the BOS area at very low HHs.

Regarding the effects of FSLs on the risk of a fall, Maki, Holliday, and Topper (1994) conducted a prospective study and determined that the lateral sway amplitude is the best predictor of future falling risk in the elderly population. Holbein et al. (2007) investigated that FSLs are positively correlated with the displacement of postural sway during static standing and a valid indicator of balance ability. Fujimoto et al. (2015) examined the COP trajectories during perturbations derived from external forces for participants both with and without the history of a fall (fallers/nonfallers). The FSLs for fallers were reduced compared to nonfallers, and the stability margin, which is the minimum distance between COP and BOS, was greater for fallers than nonfallers. The reduced FSLs predispose people to more precarious stability conditions by limiting their ability to regulate COM momentum induced by perturbations (Fujimoto et al., 2015).

At work sites, the use of the human simulation software, called the digital human model, helps to accurately estimate the risk of falls (Kawano et al., 2003, 2009). However, few models use the models of stability limits and regions considering the effects of squatting postures. Therefore, the estimated postural stability based on the FSLs for upright positions may underestimate the risk of a fall for
tasks involving squatting postures. The estimates of FSLs that vary depending on the squatting postures contribute to a more accurate estimation of the risk of a fall.

4.2 | Relation between hip-height and lower-limb loads

With a decrease in the HH, the flexion of the knee joint becomes deeper, and the activity of the tibialis anterior muscle approximately linearly increases. We evaluated the correlation between muscle activities and joint torque ratios by Pearson’s correlation coefficient. The R and L tibialis anterior muscles highly correlate with the knee-joint torque ratios (R: \( r = 0.967 \); L: \( r = 0.955 \)), while no significant correlation was observed between other joints and muscles.

During the squatting tasks, the participant needs to maintain the angles of ankle joints within an appropriate range to prevent a fall. The dorsiflexion angles of the ankles were close to the maximum range of motion (ROM) when the HH is lower than 90%. Therefore, the tibialis anterior muscles require the stronger contractions compared to the normal positions because the joint position around the maximum ROM generates a passive torque in the opposite rotational direction (Chaffin et al., 2006). Thus, the tibialis anterior muscles were significantly correlated with the knee-joint torque. However, the activity of the rectus femoris was gradually reduced when the HH was lower than 40%. These phenomena are attributable to the contact between the thighs and lower legs. When the thigh is strongly contacted with the posterior surface of the lower leg in the deepest squatting position, the applied reaction forces generate a passive torque, which offsets the joint torque on the knee joints (Fukunaga, Ayaka, Ito, & Morimoto, 2016; Zelle, Barink, Loeffen, De Waal Malefijt, & Verdonschot, 2007). Consequently, the activity of the rectus femoris differs from the trend of the knee-joint torque.

The joint torque estimation in this study was based on the inverse dynamics approach, where several factors can influence the results. Hence, only particular muscles were significantly correlated with the joint torque. We assumed the torque on body joints as the external mechanical moments around the joints, maintaining equilibrium with internal forces generated by muscles, tendons, or articular capsule. Then the angle of pull, the moment arm to the center of rotation, and even the structure of the particular muscle significantly vary during normal movements (Redfern, 1992). Therefore, sEMG values do not necessarily reflect the mechanical loads on the biomechanical system during complex and dynamic movements. To estimate the torques exerted by the muscles, the biomechanical model with a more precise inner structure is needed.

4.3 | Relevance of lower-limb loads on the functional stability limits during squatting

We now discuss the effect of hip height on the FSL based on the joint torque ratio and sEMG values. For this purpose, we divide the 11 experimental conditions into three groups. In the first group (100–70% HHs), the FSLs remained constant. In this group, the muscular activities of the rectus femoris and tibialis anterior were below 30% MVC, and the knee-joint torque ratio was at most 30%. These results indicate that while leaning and moving their hips, the participants sufficiently shifted their COP despite the increased physical loads on the lower limbs.

In the second group (60–40% HHs), the FSLs narrowed toward the inner edge of the feet. The COP shift was restricted in the left and right directions, significantly reducing the FSL area from that of upright standing. This change could be attributed to the progressively increased muscular loads on the lower limbs imposed by the deepening squatting postures. At 40% HH, the maximum knee-joint...
torque ratio approached 50%. Under this considerable load, the body is not easily supported by a single leg. Therefore, the participants probably narrowed their FSL area until their body weight was supported on both legs. This expectation is supported by the increased muscular activity of the rectus femoris, which increased from 8.7% MVC during upright standing to 49.7% MVC during squatting at 40% HH. Based on the relation between the endurance time and relative force derived by Van Dieën and Oude Vrielink (1994), the estimated endurance time of exerting a 50% MVC relative force was 1.2–3.4 min (average: 1.9 min).

In the third group (30–0% HH), FSL in the forward direction at <30% HH was lower than at 40% HH. It is thought that the forward shift of the COP is limited by the ROM in the ankle dorsiflexion. During deep squatting (30–0% HH), the participants shifted their COP horizontally by dorsiflexing their ankle joints rather than bending their knee joints. This action is attributable to the increased impedance in the knee joint as the muscle contractions enhance around those joints. This co-contraction of the muscles decreases the number of degrees of freedom of the horizontal pelvic movements (Jensen, 2008). Therefore, as the torque increased in their knee joints during deep squatting, the participants dominantly shifted their COP by moving their ankle joints. This interpretation is supported by the high muscular activity of the tibialis anterior.

4.4 Limitations

This study has several limitations. First, the FSLs of the feet positions were measured only during static squatting postures. The participants were instructed to maintain their hips at constant height while moving their pelvis in a circular motion. In any practical work environment, squatting motions are accompanied by a more complex series of movements, and some work tasks require different positions of the feet. These movements were not considered here. Further studies will examine more realistic motions on participants over a wider age range, enabling a comprehensive quantification of the FSL indicator.

As revealed in previous studies, FSLs are affected by various factors such as foot placements, anthropometric factors, age, motor control strategies (which differ among individuals), and the COP parameters in the time and frequency domains (Chiari, Rocchi, & Cappello, 2002; Fujimoto et al., 2015; Kasahara, Saito, Anjiki, & Osanai, 2015; Kilby, Slobounov, & Newell, 2014; Kirby, Price, & MacLeod, 1987). These effects need to be evaluated in further study. Especially important is the dynamic property of postural balance, which relates to the fall risk on real work sites. For the dynamic balance of a biped robot, several models have been introduced as indicators of whether the current state is balanced or falling. The prevalent approaches are based on the virtual foot-rotation point on the ground (Goswami, 1999), the ability of a biped system to come to a stop after taking maximum N steps (Koolen, De Boer, Rebula, Goswami, & Pratt, 2012), or a contact-specific partition of the COM state space (position and velocity) (Mummolo, Peng, Gonzalez, & Kim, 2018). Moreover, we may be able to apply dynamic mechanical models and evaluation indices such as virtual time-to-contact approaches, which quantify the temporal proximity to the stability boundary (Dutt-Mazumder, Challis, & Newell, 2016; Kilby et al., 2014).

5 CONCLUSIONS

The present study investigated the effect of HH on the area of the functional stability limit during squatting. Lowering the HH of the squatting posture reduced the moving ranges of the COP by up to 20%, 40% and 60% in the forward, right, and left/backward directions, respectively. In the deepest squatting posture, the FSL area decreased up to 20% of the BOS area. These trends are attributable to the difficulty of supporting the deep squatting posture by a single leg, as the muscular activities increase to over 30% MVC around the knee joints. The results can also be explained by the reduced ROM of the ankle dorsiflexion, by which the participants swayed their body, during the deepest squatting.

When evaluating postural stability during squatting, we should acknowledge the narrower FSLs during squatting than during upright standing. In other words, if inspectors assume the same FSLs in upright and squatting postures, they will likely underestimate the fall risk. Furthermore, the relevant indicators of postural stability may need to be revised for different working postures and environments with different heights.

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