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Investigation of sit-to-stand and stand-to-sit in an above knee amputee

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Abstract

the objective of this pilot study is twofold: 1) to extract key factors/features in sit-to-stand and stand-to-sit (STS) performed by an above knee (AK) amputee; 2) to propose a convenient way to quantify symmetry. One male unilateral transfemoral amputee participated in the pilot study. The subject was instructed to rise in a comfortable and natural manner and conduct a series of sit-to-stand, stand-to-sit. We simultaneously measured kinematics, kinetics and muscle activities. Principal component analysis (PCA) was used to reduce the dimension and identify modes and a convenient index of STS symmetry (slope of the major axis of the error ellipse) is proposed using the insole pressure sensors. Based on the preliminary results it is recommended that kinematics and kinetics in both the sagittal and frontal planes be considered for an AK amputee performing STS. The information might be useful for further research on amputee STS.

I. INTRODUCTION

STS movement is an important function in daily life. However, the seemingly easy task performed by healthy individuals in routine manner could potentially pose serious challenge to people with movement disorders/disabilities. Sit-to-stand is essential to initiate ambulation and determines the independence of the person. STS has been extensively studied especially in healthy subjects with various factors examined, such as age, gender, chair height, arm rest, foot placement and so on [1, 2]. However, few studies have been conducted to investigate STS in people with movement disorders and studies of STS on amputees are even less. Ambulation is an essential motor function in amputee rehabilitation and without successful STS amputees could not even take advantage of the prostheses prescribed and their independence will be compromised. To conduct STS, various muscles and joints need to coordinate in a synergistic manner. It is obvious that amputee is facing greater challenge due to the loss of muscles and joints especially for above knee amputee who losses both knee and ankle joints. Typically, compensation strategy will be used especially in unilateral amputees. For example, studies have shown that amputees rely heavily on their sound side for power generation during STS [3, 4]. Previous studies on STS are mainly focused on kinematics and kinetics and there are few studies on muscle activities. EMGs studies were primarily tested on able-bodied subjects [5, 6]. By comparing performance between able-bodied and people with movement disorders, it has been shown that the mechanical demanding is not balanced in the latter group due to the reduction and/or loss of power generation. Therefore, index of loading symmetry is one of the most popularly used measurements to quantify the performance. Both joint torques and ground reaction forces have been used and peak values are typically used to calculate the symmetry index. For example, the peak magnitude of GRF and knee/hip joint moment was used to obtain the degree of asymmetry in a study comparing the power knee and C-leg [4]. Picking peak values significantly simplifies the procedure of quantifying symmetry, however, key information related to the performance of STS might be lost by such an arbitrary selection. Most of the published studies were conducted in research lab and involved sophisticated instruments such as, force plate, motion capture system, EMGs, multi-axis load cell. However, in clinic, practitioners need to quantify performance of STS quickly and conveniently without access to those instruments. Therefore, there is a need to simplify the procedure of quantifying STS performance. A better understanding of STS in amputees, especially above-knee (AK), will help us identify key factors associated with successful STS and provide recommendation on rehabilitation and training. The objective of this pilot study is therefore twofold: 1) to extract key factors/features in STS performed by AK amputee; 2) to propose a convenient way to quantify loading symmetry.

II. Methodology

A. Subject and experimental protocol

One male unilateral AK amputee (right side is amputated; age: 57 yr; weight: 75.8kg; Height: 175.3cm; Duration: 32 yr) participated in the pilot study after giving consent form. The ratio of the length of residual limb (distance between the greater trochanter [GT] and the distal end) to the thing length (distance between the GT and lateral epicondyle) of the sound side is .51. The TFA used his prosthesis: a Manuch® SNS knee and a 27 cm Vari-Flex® foot. The check socket for the test was made by duplicating his definite socket (an ischial containment design with suction suspension). During the experiment an elastic suspension belt was also used to reinforce the socket suspension. A comfortable chair with appropriate height was selected and secured to the floor (Fig. 1). The subject was instructed to rise in a comfortable and natural manner and conduct a series of sit-to-stand, stand-to-sit, without arm assistance at self-selected rate. A practice session was given prior to the test to allow the subject to get familiar with the protocol.

B. Data acquisition

Surface EMGs—Surface EMGs were used to register the activities of the following muscles (Note: due to the surgery and altered insertion sites, the electrodes were placed approximately): Gluteus Maximus (GMax), Gluteus Medius (GMed), Rectus Femoris (RF), Vastus Lateralis (VL), Vastus Medialis (VM), Biceps Femoris Long (BFL) and Short (BFS) Heads, Semitendinosus (ST) and Adductor Magnus (AdM) [7]. The electrodes (MA-411-002, Motion Lab System Inc. Baton Rouge, LA) consisted of two contacts. The electrodes contained a preamplifier with a band-pass filter (frequency range: 20–3000 Hz; gain: 20). A specially designed adaptor (Liberating Technologies, Inc., Holliston, MA) was placed on the gel liner which makes a nice contact with the targeted muscle and an electrode clamp was used to hook up the corresponding EMG site. The skin was conditioned (shaved and cleaned) before socket donning. The ground electrode was placed around the anterior Iliac Spine. The EMGs signals were sampled at a rate of 2500 Hz.

Kinetics—A multi-axis load cell (PY-6, Bertec Corp. USA) was mounted under the knee between the knee joint and the distal pylon to measure the three dimensional forces and torques (the sensor was mounted in such a way that the +z axis was pointing downward, +x axis is pointing medially and +y axis is pointing forward). The plantar pressures of both feet were registered using Pedar-X system (Novel, Munich, Germany). The insole sensors were carefully calibrated following the manufacturer's instruction using Trublu® calibration device. The Pedar-X system was synchronized with other signals via using a reference signal. The data acquisition was performed using Novel system software.

Kinematics—Lower extremity joint kinematics of both sides (hip adduction/abduction, hip extension/flexion, hip internal rotation/external rotation, knee flexion/extension, knee internal rotation/external rotation, ankle plantar flexion/dorsiflexion and ankle eversion/inversion) were registered using a fully wearable motion analysis system, MVN (Xsens Technologies, NL). The sensors were connected to an Xbus Masters via which the kinematics data were transmitted through wireless communication at a rate of 100 Hz. A calibration was carefully conducted before the test to allow the system to identify the body alignment and body dimensions.

C. Data analysis

All the raw data were synchronized. The signals from the multi-axis load cell were converted to physical units. The insole pressure data were exported and the sum of the contact force over the entire sensor region was calculated. The raw EMG data were band-pass filtered at 20 – 500 Hz using a zero-lag eighth-order Butterworth filter to remove the low-frequency motion artifacts and high-frequency noise. The EMG signals were further low-passed to obtain linear envelope. All the other signals were low-passed using a digital zero-lag fourth-order Butterworth filter with cut-off frequency of 5 Hz. The sit-to-stand cycle was identified as from initial trunk movement to standing erect. Similarly, the stand-to-sit cycle was determined as from initial hip movement to stable hip posture (i.e. no change in hip joint position). The time of each cycle was obtained and averaged across cycles (in total 9 cycles). Principal component analysis (PCA) was used to reduce the dimension and identify modes [8]. The covariance matrix: $\mathrm{Cov}(x_i, x_j) = E\{[x_i - E(x_i)][x_j - E(x_j)]\} = \sigma_{ij}$, was obtained first (x is the time series) and then the eigenvalues (λ) and corresponding eigenvectors (ν)were determined ($\sigma_{ij}\nu = \lambda\nu$). Dominant modes which interpret the majority of the variance were identified. PC loadings, the correlation coefficients between the PCs and the original data were obtained and shown in the results. The scatter plot was fitted using ellipse with 85% confidence interval. The slope of the major axis was quantified and used as an index of symmetry. All data analyses were conducted using MATLAB (Mathworks Inc., Natick, MA, USA).

III. RESULTS

The time for sit-to-stand and stand-to-sit was 2.37±.16 and 2.37±.33 s respectively across nine repetitions.

Using PCA, we are able to reduce the dimension of modes and identify four major PCs. For example, for the first four PCs, it is clear that majority of the hip flexors and extensors were working together (Table 1). The contribution of Adductor Magnus in PC1 and Gluteus Medius in PC2 indicates the importance of maintaining balance in frontal plane during STS.

Embedded multi-axis load cell (Table 2) provided invaluable information on kinetics seen under the knee joint. The kinetics could be mainly attributed to the force along the pylon (i.e. Fz), force in the anterior-posterior direction (i.e. Fy) and moment in the sagittal plane (i.e. Mx) as shown in the loading for PC1. In addition, medial-lateral horizontal force (i.e. Fx) and moment in the frontal plane (i.e. My) have significant loading on the second PC. The third PC, though explains smaller fraction of variance, is primarily attributed to torsion (i.e. Mz).

The joint kinematics (Table 3) revealed that hip joint especially the affected side made significant contribution (i.e. PC1). For both sides, sagittal joint kinematics were mainly interpreted by the hip and knee joints. In addition, a likely compensation between the frontal motion of the affected hip joint and the frontal motion of the sound side ankle joint is shown. The second PC is significantly influenced by sagittal ankle kinematics. The third PC, though interprets smaller fraction of the variances, could be attributed to the kinematics of sound side hip rotation in the transverse plane.

Fig. 3 illustrates the profiles of Fzs registered directly using the multi-axis load cell and obtained via Pedar insole pressure sensors. Though there is some deviation in between, in general, the two profiles are strongly correlated with a correlation coefficient of .88.

Fig. 4 illustrates the scatter plot of the summed forces for both left and right sides. The slope of the major axis of the error ellipse fitted over all the collected data points (red dots) is 0.0747. The symmetry as characterized as load bearing will be improved as the slopes increases and reaches perfect when the slope is one.

IV. Discussion and conclusion

In this pilot study, we have simultaneously measured kinematics, kinetics and muscle activities in STS performed by an AK amputee. The averaged sit-to-stand time, 2.37±.16s, is slightly shorter than that reported in a previous study, 2.62±.87s. This might be attributed to the higher level activity of the subject. It is clear that the information is redundant and PCA has been used to reduce the number of modes (i.e. PCs). The major loadings on the muscle activities are hip and knee flexors and extensors. In addition, the contribution of adductor and abductor should not be ignored during amputee STS due to their importance in stabilizing in the frontal plane. Besides the importance of kinematics in the sagittal plane, there is a likely compensation between the sound side and the affected side in the frontal plane as evidenced as significant loadings of sound side ankle eversion/inversion and affected side hip joint abduction/adduction. Furthermore, the hip internal/external rotation on the sound side explains smaller yet still significant variance in kinematics.

In this pilot study, multi-axis load cell was used to register the torque and force seen under the knee joint. Force along the pylon and the moment in the sagittal plane are the two major loadings on the first PC. Meanwhile, these two major players in the sagittal plane might be captured by the insole pressure sensor as shown in fig. 3. The force normal to the contact surface of the insole might be decomposed into two components with one along the pylon (similar to Fz measured using PY-6) and another perpendicular to the pylon (will cause moment in the sagittal plane). The significance of stabilization in the frontal plane is revealed in the EMGs. Similarly, the significance is highlighted in the kinetics measured with the multi-axis load cell and both Fx and My are contributing significantly to the second PC. The third PC is mainly due to the hip torsion and relates to the hip joint internal/external rotation as revealed in the joint kinematics. The strong association between the measurements of PY-6 and Pedar insole might indicate that insole pressure sensor could be a useful and convenient tool in the clinic in case multi-axis load cell is not accessible and/or not possible to be used. The joint kinematics patterns are related to those seen in EMGs. In addition, it is important to also take a look at the ankle kinematics especially in the sagittal plane.

A convenient index of STS symmetry (slope of the major axis of the error ellipse) is proposed. Unlike previous studies in which only peak values were used, all the data points registered were utilized. The weight-bearing (i.e. the summation of the force normal to the insole) between legs will show more or less symmetric pattern in able-bodied person. In other words, if illustrated in a scatter plot the slope of the major axis of the error ellipse will be around one. In the current study, it is clearly shown that the slope is fairly shallow due to the significant imbalance between legs. This index is recommended in that it takes all the measurements into account and it will be not biased at least by arbitrarily selecting points (e.g. the peak values).

Limitations of the study are acknowledged. Only one AK amputee was tested and more subjects will be recruited in future study. EMGs were only tested on the affected side and it will be interesting to compare the muscle activities between sound and affected sides. Joints moments were typically used to quantify STS symmetry. However, a recent study [9] has shown a discrepancy between the torque values measured directly and those obtained via inverse dynamics. It might be cautious to use the joint moments obtained via inverse dynamic approach especially for AK amputee STS in which the frontal stabilization is also crucial. Upper body motion might also be important in performing STS and will be included in the future. In addition, it has been suggested that STS should involve practicing standing up at a range of speeds since the ability to control speed variation is critical for independence. [10].

In summary, in the pilot study, key factors and/or features were identified and a convenient index of STS loading symmetry was proposed. It is recommended that kinematics and kinetics in both the sagittal and frontal planes be considered for an AK amputee performing STS. The information might be useful for further research on amputee STS.

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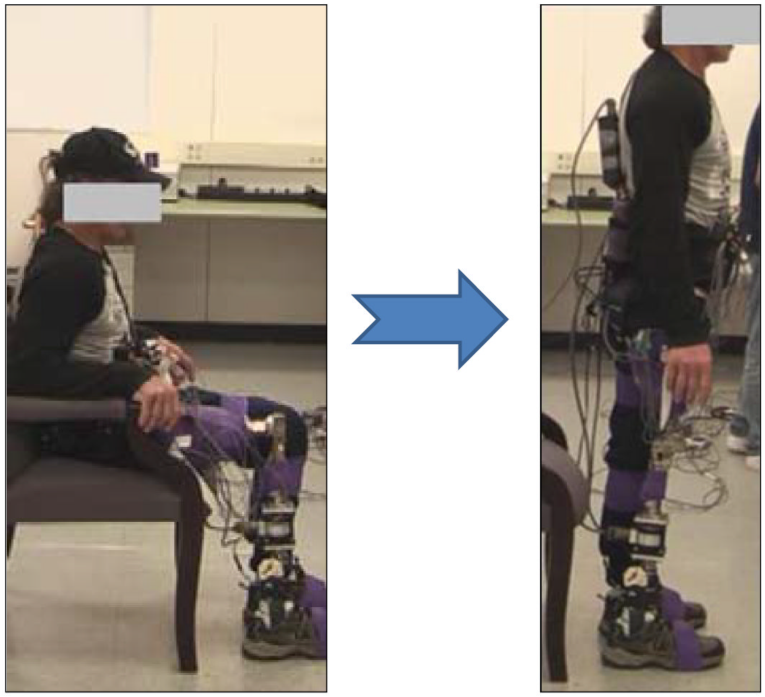

Fig. 1.

Experimental setup of sit-to-stand, stand-to-sit. Left panel: initial posture; Right panel: standing erect.

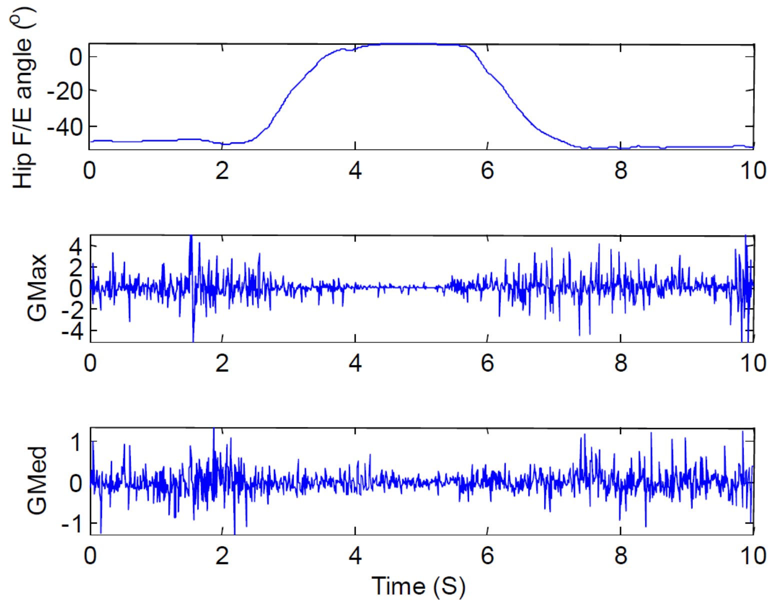

Fig. 2.
Hip joint kinematics and muscle activities during one complete sit-to-stand and stand-to-sit cycle. Note: F/E is flexion/extension.

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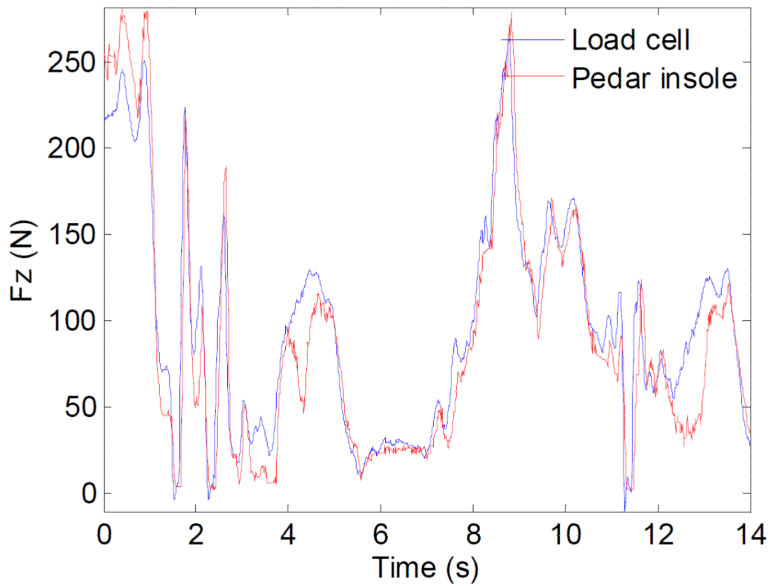

Fig. 3.
Fz (multi-axis load cell) vs. Fz (insole)

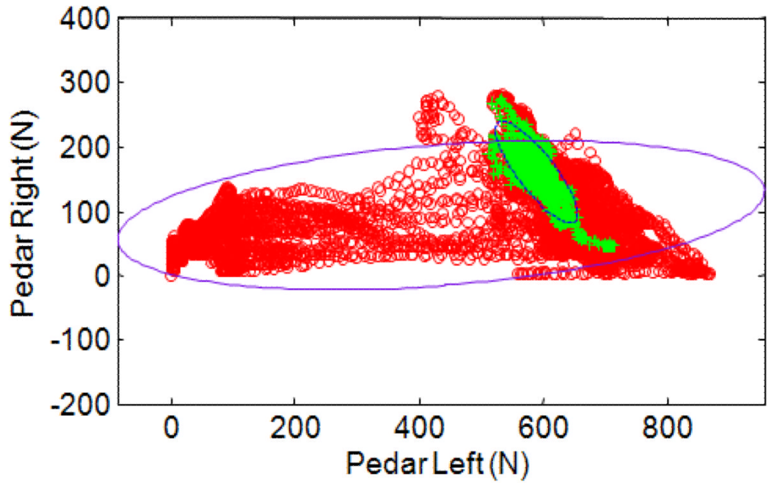

Fig. 4.
Scatter plot of Fz (Left v.s. Right). The green dots are registered during the stable standing and the slope of the error ellipse (the small one inside the green dots) is about −1. The purple ellipse is fitted over all the data points.

Table 1

PCs of surface EMGs

Muscles	PC1	PC2	PC3	PC4
GMax	0.2722	**0.8718**	0.2716	0.0746
GMed	0.4556	**0.7620**	0.3006	−0.1266
VM	0.1343	−0.2040	**0.6398**	−.3871
RF	**0.6690**	−0.3681	0.0150	−0.1583
VL	**0.7051**	−0.3531	0.1570	−0.2722
BFL	**0.7950**	0.1065	−0.4518	−0.0698
BFS	0.3980	−0.3192	**0.6163**	0.2987
ST	0.4228	−0.1237	0.1113	**0.7970**
AdM	**0.8538**	0.0384	−0.4108	−0.0154
Var (%)	*32.70*	*19.69*	*15.15*	*11.11*

Note: major loadings are highlighted with underlines.

Table 2

PCs of Kinetics (PY-6 multi-axis load cell)

	PC1	PC2	PC3
Fx	0.4663	**0.7837**	0.2279
Fy	**0.7318**	−0.1163	0.5843
Fz	**0.8157**	−0.0177	−0.3279
Mx	**0.9178**	−0.0782	0.1862
My	−0.4132	**0.8350**	0.1044
Mz	0.5609	0.2691	**−0.7026**
Var (%)	*45.76*	*23.40*	*17.33*

Table 3

PCs of Joint Kinematics

Joint Kinematics	PC1	PC2	PC3
R-Hip Adduction(+)/Abduction (−)	**−0.9213**	−0.2692	0.0904
R-Hip Extension(+)/Flexion(−)	**−0.9669**	0.0690	0.0469
R-Hip Internal Rot(+)/ External Rot(−)	**0.9149**	−0.1777	0.1103
R-Knee Flexion (+)/Extension(−)	**0.9624**	−0.1948	−0.0545
R-Knee Internal Rot(+)/ External Rot(−)	**0.8278**	0.1962	−0.2432
R-Ankle Plantar (+)/Dorsi(−)	−0.3695	**0.7744**	0.2086
R-Ankle Eversion (+)/inversion (−)	0.5969	0.2227	0.2540
L-Hip Adduction(+)/Abduction (−)	0.4197	−0.5720	0.1067
L-Hip Extension(+)/Flexion(−)	**−0.9586**	0.0617	0.0547
L-Hip Internal Rot(+)/ External Rot(−)	−0.1308	0.5316	**−.0.7715**
L-Knee Flexion (+)/Extension(−)	**0.9477**	−0.1678	0.0092
L-Knee Internal Rot(+)/ External Rot(−)	**−0.9117**	−0.0397	0.2567
L-Ankle Plantar (+)/Dorsi(−)	0.4833	**0.6325**	0.4216
L-Ankle Eversion (+)/inversion (−)	**−0.8025**	−0.4306	−0.1685
Var (%)	60.31	14.74	7.67