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Original Article

Development of an ankle torque measurement device for measuring ankle torque during walking

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Abstract. [Purpose] To develop a device for measuring the torque of an ankle joint during walking in order to quantify the characteristics of spasticity of the ankle and to verify the functionality of the device by testing it on the gait of an able-bodied individual and an equinovarus patient. [Subjects and Methods] An adjustable posterior strut (APS) ankle-foot orthosis (AFO) was used in which two torque sensors were mounted on the aluminum strut for measuring the anterior-posterior (AP) and medial-lateral (ML) directions. Two switches were also mounted at the heel and toe in order to detect the gait phase. An able-bodied individual and a left hemiplegic patient with equinovarus participated. They wore the device and walked on a treadmill to investigate the device's functionality. [Results] Linear relationships between the torques and the corresponding output of the torque sensors were observed. Upon the analyses of gait of an able-body subject and a hemiplegic patient, we observed toque matrices in both AP and ML directions during the gait of the both subjects. [Conclusion] We developed a device capable of measuring the torque in the AP and ML directions of ankle joints during gait.
Key words: Torque measurement device, Ankle joint, Gait phase

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INTRODUCTION

Spasticity after stroke is an indicative sign of damage to the upper motor neuron system and the cerebral cortex. In 1980, Lance defined spasticity as a velocity-dependent increase in muscle tone with exaggerated tendon jerks, resulting in hyperactive muscle activity[1)]. Clinical assessments of spasticity should include quantification through validated scoring systems such as the Ashworth scale[2)], modified Ashworth scale[3)] (MAS), Tardieu scale[4)], and modified Tardieu scale[5)]. Fonseca et al. used a portable spasticity measurement device and reported that there was no statistical difference between MAS and stiffness measured by the device[6)]. Tomita et al. reported on novel equipment for ankle stiffness measurement for clinical use[7)]. They decomposed the stiffness of the paralyzed foot ankle of a hemiplegic patient into different components.

Although the above-mentioned quantification methods have been used frequently in clinical situations, patients have to rest on a bed or in a chair during the assessment. However, some patients do not develop spasticity when they are at rest, instead they develop and have spasticity when they begin to walk. To measure spastic stiffness during walking, we developed a novel device. We verified the device's functionality by testing it on an able-bodied individual and an equinovarus patient after stroke.

SUBJECTS AND METHODS

An adjustable posterior strut ankle-foot orthosis (APS-AFO) was used. A duralumin strut, 5 mm thick, 20 mm wide, and 280–345 mm long, was attached to the foot portion of the patients' APS-AFO (TAPS-AFO, TIMS Adjustable Posterior Strut, Tomei Brace, Aichi, Japan; and RAPS-AFO, Remodeled Adjustable Posterior Strut, Tomei Brace, Aichi, Japan). A semicircular holder was attached to the other side of the strut. The length of the strut was adjusted (280–345 mm) according to the length of the patients' lower leg (Fig. 1). The angle of the ankle joint of the APS-AFO was set at 5°dorsal flexion. To measure torque in the anterior-posterior (AP) and medial-lateral (ML) directions, two torque sensors were attached to the strut with eight strain gauges configured into two full-bridges. The output of the torque sensors was transferred to a PC (SOTEC WA5512, Intel Core^{TM} 2 Duo, Processor T5500 [1.66 GHz] Windows Vista^{TM} Home Premium Redmond, WA, USA) via a sensor interface (PDC-300B, Kyowa, Japan). Two mechanical switches were placed on the bottom of the AFO at the heel

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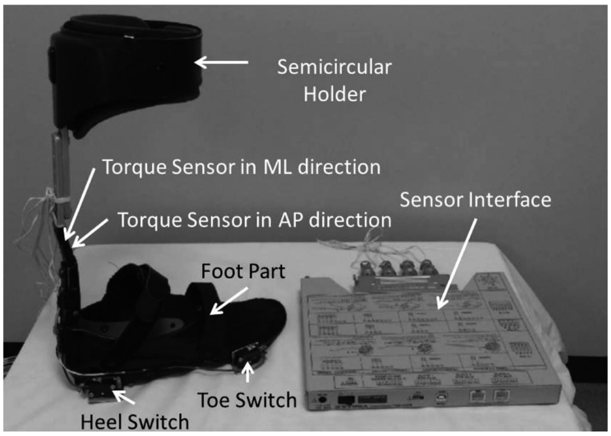

Fig. 1. Torque sensor mounted on an adjustable posterior strut ankle-foot orthosis and sensor interface
ML: medial-lateral; AP: anterior-posterior

and toe in order to define the gait phase.

Because of the geometric non-uniformity of the metal strut and asymmetric adherence of the strain gauges, bending in the AP direction interfered with the output of the torque sensor monitoring in the ML direction and vice versa. Thus, interference-free calibration was needed (equation [1]). Calibrated torque (true torque) was calculated by multiplying the measured strain with the known calibration matrix.

$$\text{Calibrated torque} = \text{measured strain} \times \text{calibration matrix} \quad (1)$$

The calibration matrix was obtained by using equation (2). We determined the applied torque, which was calculated from the product of different sand weights (−20, −15, −10, −5, −1, 0, 1, 5, 10, 15, and 20 kg), length of the strut (0.15 m), and the acceleration due to gravity (9.8 m/s^2). The outputs of the torque sensors at each applied torque were used for the measured strain, and then, we obtained the left side and the first member of the right side by using equation (2).

$$\text{Applied torque} = \text{measured strain} \times \text{calibration matrix} + \text{error} \quad (2)$$

The calibration matrix was calculated using the least squares method, which minimizes the sum of the squares of the error. The result is shown in equation (3).

$$\text{Calibration matrix} = ([\text{measured strain}]^{T} \times [\text{measured strain}])^{-1} \times [\text{measured strain}]^{T} \times [\text{applied torque}] \quad (3),$$

where the notation $()^{-1}$ and $[\ ]^{T}$ denote the inverse and transposed matrices, respectively.

The subjects of our study were a 34 years old able-bodied male subjects and a 30 years old equinovarus left hemiplegic female patient (with right putaminal hemorrhage (intracranial hemorrhage)). Both subjects walked on a treadmill at the speed of 1.2 km/h.

This study and experiment procedure were approved by the local Ethics Committee of Nanakuri Sanatorium at Fujita Health University (approval no. 97). Informed consent was obtained from both subjects. All personal information was removed to protect the subjects' identity.

RESULTS

AP and ML directions torques in Nm are shown in the 1st and 2nd columns of equation (4), respectively.

$$[\text{Applied Torque}] = \begin{bmatrix} -29.4 & 0 \\ -22.5 & 0 \\ -14.7 & 0 \\ -7.35 & 0 \\ -1.47 & 0 \\ 0 & 0 \\ 1.47 & 0 \\ 7.35 & 0 \\ 14.7 & 0 \\ 22.05 & 0 \\ 29.4 & 0 \\ 0 & -29.4 \\ 0 & -22.5 \\ 0 & -14.7 \\ 0 & -7.35 \\ 0 & -1.47 \\ 0 & 0 \\ 0 & 1.47 \\ 0 & 7.35 \\ 0 & 14.7 \\ 0 & 22.05 \\ 0 & 29.4 \end{bmatrix} \quad (4)$$

The corresponding strains in με are shown in the 1st and 2nd coloumns of equation (5), respectively.

$$[\text{Measured strain}] = \begin{bmatrix} 11000 & -138 \\ 8540 & -107 \\ 5850 & -71 \\ 2940 & -37 \\ 570 & -8 \\ 0 & 0 \\ -594 & 12 \\ -2900 & 38 \\ -5700 & 71 \\ -8720 & 109 \\ -11200 & 134 \\ 38 & 272 \\ 35 & 204 \\ 30 & 137 \\ 24 & 69 \\ 14 & 15 \\ 0 & 0 \\ -14 & -12 \\ -10 & -68 \\ -30 & -138 \\ -26 & -206 \\ -30 & -274 \end{bmatrix} \quad (5)$$

By substituting the equations (4) and (5) into the equation (3), the calibration matrix was obtained as follows.

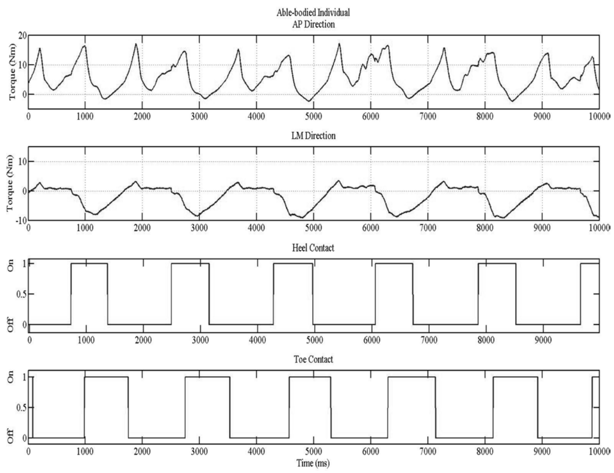

Fig. 2. Torques of an able-bodied individual in the anterior-posterior (AP) and medial-lateral (ML) directions

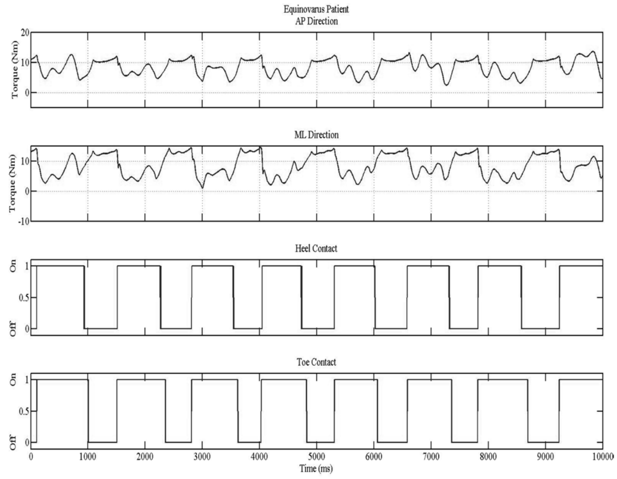

Fig. 3. Torques of an equinovarus patient in the anterior-posterior (AP) and medial-lateral (ML) directions

$$\begin{bmatrix} -0.0012 & 0.0006 \\ -0.0019 & -0.0498 \end{bmatrix} \quad (6)$$

To understand the characteristics of the device, we applied the torques of the AP direction alone by using sand weights, and we obtained the calibrated torque, which was derived from the calibration matrix. In the AP direction, deviation from the expected value was 0.39 Nm on average and was 0.62 Nm at the maximum. The linearity, defined by the ratio of the average deviation to the full scale, was 1.3%. In the ML direction, deviation from the abscissa was 0.25 Nm on average and was 0.48 Nm at maximum. Linearity was 0.85%.

We applied the torques of the ML direction alone by using sand weights and obtained the calibrated torque, derived from the calibration matrix. In the ML direction, deviation from the expected value was 0.024 Nm on average and was 0.031 Nm at the maximum. Linearity was 0.082%. In the AP direction, deviation from the abscissa was 0.12 Nm on average and was 0.16 Nm at the maximum. Linearity was 0.41%.

The torque and heel toe motion of an individual with a normal gait is shown in Fig. 2. The torque in the AP and ML directions, the toe switch status, and the heel switch status are shown. When the switches were on, it indicated that the sole was in contact with the floor, and the line was in the upper position, however when they were off, the line was in the lower position. First, the heel contacted the floor (the belt of the treadmill, initial contact phase), and then the heel and toe made contact with the floor (mid-stance phase). This was followed by releasing the heel and leaving the toe in contact with the floor (late stance phase). Finally, both switches were off, indicating that the foot left the floor (swing phase).

The peak of the plantar flexion torque occurred immediately after heel contact (the first half of the load reaction phase) and then it gradually decreased towards the mid-stance phase. In the ML direction, eversion torque was at its maximum after the heel contact, which decreased gradually and extinguished in the swing phase. Almost no inversion torque was observed in the subject with normal gait.

The gait of the equinovarus patient is shown in Fig. 3. The heel and toe made contact at almost at the same time, with the subject showing a flat foot during initial contact with the floor. During all phases, plantar flexion torque was observed in the AP direction and inversion torque was observed in the ML direction. In the late swing phase a large planter flexion and inversion torque was observed, typically a result of spastic muscle hypertonia.

DISCUSSION

Calibrated torque in the AP direction was approximately proportional to the applied torque and the calibrated torque in the ML direction was negligible. Similarly, calibrated torque in the ML direction was approximately proportional to the applied torque and calibrated torque in the AP direction was negligible.

Interference between the torque in the AP and ML directions was sufficiently compensated by the calibration matrix, which achieved our requirement.

The observed torque in normal gait shows a small plantar flexion torque at initial contact and the first half of the load reaction phase which decreases towards the mid-stance phase. These findings agree with those of Perry[8)]

The equinovarus gait had planter flexion, and inversion torques was observed in all phases. In the stance phase, torque around the ankle joint is caused by body weight, floor reaction force, and muscle contractions; thus, it is difficult to separate these individual torques. In the swing phase, the torque caused by the body weight and floor reaction force disappears and the torque caused by the muscle contraction alone can be observed. Torque observed in the swing phase of the equinovarus patient's ankle joint, which was much larger than the torque observed in the swing phase of the able-bodied individual, must be a result of an increase in muscle tone (spastic contraction).

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