Abstract

The authors proposed an electromyography computed tomography (EMG–CT) method to measure the distribution of muscle activity in the forearm using surface EMG signals from multiple surface electrodes. The present study is to develop a method to estimate muscle stress, i.e., force generated during contraction per unit area in the whole cross-section of the forearm based on EMG–CT. While three subjects performed hand gripping trials using three hand grip devices, EMG signals around the forearm were measured using EMG–CT. An EMG conduction model of the forearm was constructed using an outline geometry of the subjects’ forearm which was measured with a handy 3D scanner. The stress of muscle was calculated from the relationship between gripping force and total muscle activity. As a result, the distribution of muscle stress in the forearm during hand gripping was visualized in a tomographic image. It was clear that the stress was concentrated in the flexor digitorum superficialis, flexor digitorum profundus, flexor carpi radialis, and extensor digitorum communis region. The maximum stress in the forearm muscles increased from 0.08 ± 0.01 to 0.18 ± 0.02 MPa when gripping force increased from 77 to 242 N. This study provides a novel method of measuring muscle stress in forearm.

Key words: Biomechanics, Muscle, Electromyography, Muscle force, Forearm

1. Introduction

Hand gripping is an essential function used in many activities of daily living and working such as using a hand tool, carrying an object, and turning a doorknob. Hand gripping requires forces exerted by the fingers and the thumb, which are mainly controlled by the muscles of the forearm; thus, the gripping abilities of the hand are directly related to the strength of the forearm muscles. Measuring muscle stress defined as force generated in the muscles per unit cross-sectional area of the forearm is of great importance because it could help to understand how internal load is shared within the forearm muscles and to provide more insight into muscle mechanics.

Some researchers have attempted to measure muscle force using force transducers implanted into the forearm (Schuind, et al., 1992; Dennerlein, et al., 1998). Direct measurement of forces in forearm muscle is impractical and invasive. The muscle function and condition can be determined by electromyography (EMG) signals. Surface electrode is usually used to measure EMG signal from skin surface due to its non-invasive and ease of use. Many studies related EMG signals to muscle force generation (Buchanan, et al., 1993; Duque, et al., 1995; Hoozemans and van Dieen, 2005; Vigouroux, et al., 2007; Disselhorst-Klug, et al., 2009). However, conventional surface EMG cannot assess individual muscle activity due to high distortion of signals from noise generated by nearby muscles, making it difficult to estimate stress distribution within the forearm muscles.

In previous study, the authors proposed an electromyography computed tomography (EMG–CT) method to measure the distribution of muscle activity in the forearm (Nakajima, et al., 2014). The muscle activities in the whole
cross-section were calculated from EMG signals obtained by multiple surface electrodes attached around the forearm. It was demonstrated that the EMG–CT method can measure muscle activity distribution in the whole cross-section of the forearm in a non-invasive manner. The muscle activity distribution identified by EMG–CT has high potential to estimate muscle stress in the forearm muscles.

Therefore, this study is to propose a novel method to estimate muscle stress generated in the forearm during hand gripping, using muscle activity data obtained by EMG–CT. For calculation of stress, the cross-sectional geometry of the forearm should be measured. Recently, fast and low-cost handy 3D scanning technologies have been increasingly accepted as an efficient approach to collecting body segment data (Stančić, et al., 2013; Van den Herrewegen, et al., 2014). A forearm model was constructed from subjects’ forearm geometry obtained with a handy 3D scanner. The stress distribution within the forearm during gripping loads was estimated and visualized in tomographic images. The developed method has many practical applications such as determining the effectiveness of surgical or rehabilitation procedures by monitoring muscle recovery progress, or used as a diagnostic tool for detecting diseased or injured muscle within the patient forearm so it can be treated more accurately.

2. Methods

2.1 Experimental setup

Three male subjects (age, 23 ± 0 years; height, 169.7 ± 4.5 cm; weight, 65 ± 5.0 kg; mean ± SD) participated in this study. The subjects sat on a chair with their dominant arm placed on a horizontal table. The upper arm was at approximately 0° of abduction, the elbow joint was flexed at approximately 90°, and the wrist was placed in supine position (Fig. 1). EMG signals from the forearm were recorded using an EMG band (Nakajima, et al., 2014), consisting of 20 electrode plates. The middle point of the EMG band was positioned at the middle point of the forearm lengthwise, between the lateral epicondyle of the humerus and the radial styloid. Before attachment of the EMG band, the subject’s forearm skin was cleaned with an alcohol swab. Fat and skin thickness were measured using a skinfold caliper (Marutech, Japan).

The subjects performed gripping trials using three hand grip devices as shown in Figure 2. Maximum gripping force \( F \) was defined as the force required by the subject’s hand to keep a hand grip at full grip position, as described by the following equation:

\[
F \equiv \int_{x_1}^{x_2} wd\hspace{1em}x
\]

where \( w \) is the force distributed along the handle of the hand grip during full grip position. \( x_1 \) and \( x_2 \) are the distances from the center of rotation of the hand grip to the respective ends of the handle. In static equilibrium, torque generated by the hand grip is equal to torque exerted by \( w \), as described by the following equation:

\[
T_{\text{max}} = \int_{x_1}^{x_2} wxd\hspace{1em}x
\]

where \( T_{\text{max}} \) is the maximum torque generated by the hand grip at the full grip position. It was assumed that at the full grip position, the subject’s hand exerted a uniformly distributed force (constant \( w \)) on the handle of the hand grip. Thus, \( F \) can be calculated by the following equation:

\[
F = \frac{2T_{\text{max}}}{(x_2 + x_1)}
\]
Table 1 shows the specifications of three hand grips used in the experiment. For each gripping trial, the subject held a hand grip between the middle phalanges of fingers and palm and then squeezed the handles to full grip position and held for 5 s, three trials per load with 5 s of relaxation between trials. EMG signals were recorded during each gripping trial and relaxation using a custom program (LabVIEW 8.5, National Instruments, TX, USA). The procedures were approved by the Ethical Review Board for the Protection of Persons in the Biomedical Research Graduate School of Engineering, Hokkaido University, and all subjects signed an informed consent agreement.
Table 1. Specification of the three hand grips shown in Figure 2. $k$ is the measured torsion spring constant, $T_{\text{max}}$ is the measured maximum torque generated by the hand grip at the full grip position, and $F$ is maximum gripping force as defined in Eq. (1).

| No. | Hand grip | $k$ (N·m/degree) | $T_{\text{max}}$ (N·m) | $F$ (N) |
|-----|-----------|------------------|------------------------|--------|
| A   |           | 0.21             | 5.01                   | 77     |
| B   |           | 0.37             | 8.06                   | 124    |
| C   |           | 0.61             | 15.73                  | 242    |

2.2 Forearm model construction

To construct a forearm model of each subject, an outline of the forearm cross-sectional area was obtained according to the subject’s forearm geometry. The subject’s arm was scanned with a 3D scanner (Sense, 3D System Inc., SC, USA). During measurements, the subject sat on a chair with his arm extended away from the torso in supine position, the same position as in the gripping trial. An examiner held the 3D scanner and moved it around the subject to scan the whole arm. The scan data points of subject’s arm obtained from the 3D scanner numbered approximately 130,000 points at a resolution of a point-to-point space of approximately 0.6 mm.

The subject’s forearm geometry was reconstructed from scan data points (Fig. 3a). Scan noise was removed, and the cleaned data were exported into MATLAB (version R2014a, Mathwork, USA) for further processing. The forearm axis was drawn by connecting points between two anatomical landmarks: the lateral epicondyle of the humerus and the radial styloid. The points within ± 2-mm interval from EMG–CT analyzed section which perpendiculars to the forearm axis were projected to a plane. An outline of the forearm cross-sectional area was thus created by connecting the points using a simple convex polygonal approximation method (Fig. 3b).

An EMG conduction model was constructed using the outline of the forearm cross-sectional area. Muscle element nodes were distributed across the entire cross-sectional with an element size of 1 mm at the surface and 5 mm for the inside region. The area of each muscle element was divided using Voronoi tessellation (Nakajima, et al., 2014). The surface electrodes were placed around the outline of the forearm model. The conduction distance of each muscle element was the distance between the muscle element and the surface electrode (Fig. 3c).

2.3 Muscle activity calculation

The attenuation of the action potential depends on tissue conductivity and the distance between a muscle element and the surface electrode (Nakajima, et al., 2014). In this study, we considered that each muscle element was activated independently. Thus, the statistical summation of power of the EMG signals was possible. The mean square value of EMG from all muscle elements $i$ detected by bipolar electrodes $j$, $V_j$ can be simply expressed by the following equation:

$$
\bar{V}_j^2 = V_0(d_j)^b \sum_i m_i^2 \left( \frac{l_{ij}}{l_0} \right)^{2b(d_j)}
$$

where $V_0(d_j)$ (mV·s/(mA dipole)) is a transformation coefficient that depends on the distance between the pair of bipolar electrodes $d_j$, $m_i$ (mA dipole/s) is the muscle activity of the element $i$, $l_{ij}$ (mm) is the conduction distance between the muscle element $i$ and a pair of bipolar electrodes $j$, $l_0$ is the unit length (1 mm), and $b(d_j)$ is the power exponent of the attenuation under the distance $d_j$. In this study, $d_j$ was 15 and 45 mm. $V_0$ and $b$ were 162 mV·s/mA dipole and -2.12 at $d_j = 15$ mm and 115 mV·s/mA dipole and -1.74 at $d_j = 45$ mm (Nakajima, et al., 2014).

The activity of each muscle element $m_i$ was calculated using a sequential quadratic programming method to minimize the objective function (OF), which was defined as the sum of the power of the differences between the measured EMG $V_{Mj}$ and the calculated EMG $V_j$, as shown by the following equation:

$$
\text{OF} = \sum_j (V_{Mj} - V_j)^2
$$
2.4 Stress calculation

When external forces are applied to the fingers, the muscles of the forearm generate reaction forces to maintain a static equilibrium. Total muscle activity $\Sigma m_i$ is defined as the sum of muscle activity within the forearm cross-sectional area and reflects the amount of force generated by the muscles during contraction. A linear relationship between $F$ and $\Sigma m_i$ was assumed in this study. The muscle activity-force coefficient $\alpha$ (N·s/mA dipole) was defined as shown by the following equation:

$$F = \alpha \sum_i m_i$$  \hspace{1cm} (6)

Here, the muscle force generated by each element can be calculated from the stress within each element multiplied by its area, as shown by the following equation:

$$f_i = \sigma_i a_i$$  \hspace{1cm} (7)

where $f_i$ (N) is the force generated by muscle element $i$, $\sigma_i$ (MPa) is the stress in muscle element $i$, and $a_i$ (mm$^2$) is the area of a muscle element $i$. Thus, $F$ is equal to the sum of the force generated by all elements within the forearm and can be expressed by the following equation:
\[ F = \sum_{i} f_i = \sum_{i} \sigma_i a_i = \alpha \sum_{i} m_i \]  

(8)

The stress of each element can be calculated from the following equation:

\[ \sigma_i = \frac{\alpha m_i}{a_i} \]  

(9)

Thus, the stress distribution in the forearm muscles can be calculated using muscle activity measured by EMG–CT and Eq. (9).

3. Results

Figure 4a shows an example of EMG–CT result of subject A for the first trial using 124 N of gripping force. The distribution of muscle activity can be observed. The intensity measured for each muscle element represents the level of muscle activity. The result provides an outline of the subject’s forearm geometry in a supine position. The palmar side is represented at the upper side of the image, and the radial side can be found at the right-hand side of the image.

Figure 5 shows the relationship between \( F \) and the total muscle activity within the forearm of all subjects. Total muscle activities were plotted against gripping force, and the relationship was described by a linear regression \( (R^2 = 0.97 \pm 0.04) \). The mean of the total muscle activity of all subjects increased from 1626 ± 344 to 3878 ± 170 mA dipole/s when the gripping force increased from 77 to 242 N. \( \alpha \) of subjects A, B, and C were 0.054, 0.060, and 0.062 N·s/mA dipole, respectively.

An example of stress distribution within the forearm muscle of subject A was calculated from \( \alpha \) and Eq. (9) as shown in Figure 4. During gripping, muscle activity was distributed across the entire cross-sectional area, indicating the cooperative activity of forearm muscles (Fig. 4a). Stress was calculated by the method described and is shown in Figure 4b. Stress distribution showed a trend consistent with the muscle activity pattern, although significant stress at the surface region was detected. The position of maximum activity is different from that of maximum muscle stress, because stress calculation considers the area of muscle elements unlike muscle activity calculation.

Fig. 4  (a) Typical muscle activity distribution of subject A during 124 N of gripping force. (b) The stress distribution within the forearm was calculated from the muscle activity distribution as shown in (a).
Figure 6 shows tomographic images representing stress distribution within the forearm muscles during gripping in all subjects. Patterns of stress distribution could be observed under all investigated conditions. The results showed a change in level and area for all conditions, and an increase in stress concomitant with load increase was observed. A magnetic resonance image (MRI) of the forearm cross-sectional area of each subject is shown in the top row for anatomical comparison (Fig. 6a). The activated area and the maximum value of muscle stress increased with gripping force in all subjects. Under the 77 and 124 N load conditions, muscles in ulnar region and radial–dorsal region were active in all subjects. Under the 242 N load condition, muscles in the palmar region were active in subjects A and B. The average maximum stresses in all subjects under 77, 124, and 242 N load conditions were 0.08 ± 0.01, 0.11 ± 0.01, and 0.18 ± 0.02 MPa, respectively. In subject A, the maximum stress was found in the ulnar region under all load conditions, whereas in subject B and C, the maximum stress was found in the radial–dorsal region under all load conditions.

4. Discussion

We demonstrated that the stress distribution within the forearm muscles during gripping can be estimated and presented by the EMG–CT method. We constructed an EMG conduction model based on subjects’ forearm geometry. We developed a mathematical model relating muscle activity and force to calculate stress distribution. To the best of our knowledge, this is the first method that allows measuring stress distribution within the forearm muscles in a non-invasive manner.

In this study, we investigated the relationship between external force and muscle activity during gripping. We used a linear relationship between total muscle activity within the forearm and $F$ at various gripping forces to estimate $\alpha$ of each subject (Fig. 5). The muscle force required to maintain a static equilibrium increased with the gripping force. Generally, muscles increase their force output by recruiting more motor units or by increasing the muscle firing rate, resulting in an increase in EMG signals. Nakajima et al. (2014) noted that muscle activity during finger motion increases with the external load. The results from the present study agree with the previously reported trends. Many studies (Messier, et al., 1971; Pruijm, et al., 1980; Hof, 1984; Karlsson and Gerdle, 2001; Del Santo, et al., 2007) reported a linear relationship between EMG and external force (torque) under isometric conditions. A linear model appears to offer a good approximation of the relationship between muscle activity and force under isometric conditions.
Stress levels in forearm muscles of subjects can be measured by the method we have described (Fig. 6). Stress in each muscle element did not increase linearly with load. More muscles were active when load increased. It appears that the human hand has a mechanism that distributes load between muscles so that the stress is not concentrated in only one portion. In previous studies, values for maximum muscle stress of individual muscle fibers in mammals were measured and found to vary between 0.06 and 0.38 MPa (Close, 1969; Burke and Tsairis, 1973; Lannergren and Westerblad, 1987; Kanda and Hashizume, 1992; Buchanan, 1995). The results obtained in our study were within the range of the reported data.

The pattern of stress distribution within the forearm muscles during gripping is shown in Figure 6. Under 77 and 124 N gripping load conditions (Fig. 6b and c), the stress is concentrated in two muscle groups: on the ulnar side, which may include the flexor digitorum superficialis (FDS) and the flexor digitorum profundus (FDP), and on the radial–dorsal side, which may include the extensor digitorum communis. Subject A appeared to use more muscles during gripping, resulting in less stress concentration than the other subjects. Under the 242 N load condition (Fig. 6d), stress was distributed across the whole area. Additional stress concentration was found in subjects A and B on the palmar side which may include the flexor carpi radialis. The basic gripping function involves finger flexion generated by muscle forces from the finger flexor muscles in the forearm. This mechanism is shown by the finding that both the FDS and FDP generate stress during gripping. It is also consistent with results of previous studies that used wire

![Fig. 6 (a) Magnetic resonance image of forearm cross-sectional area. Stress distribution generated in the forearm muscle during gripping (b) 77 N, (c) 124 N, and (d) 242 N.](image-url)
electrodes to study finger motion, finding that the FDP is responsible for synchronous flexion of finger joints (Johanson, et al., 1990; Darling, et al., 1994). The maximum stress was found in the extensor region of subject B and C under the 242 N load condition (Fig. 6d). The stress appearing in the extensor might be explained by the co-contraction of the muscles to counteract the wrist flexion torque caused by the finger flexor. Many studies also found strong surface EMG signals from the extensor muscle during gripping (Hagg and Milerad, 1997; Johanson, et al., 1998; Hoozemans and van Dieen, 2005). It appears that extensors play an important role under high gripping load.

The method described here provides both quantitative values of stress levels and a distribution pattern in a non-invasive manner. This information is very useful for medical diagnosis, given that muscle force generation can be affected by muscular diseases or injury, resulting in abnormalities in stress level and distribution pattern. For example, in a patient with muscle pain, the pattern of muscle distribution might change to avoid excessive stress on the painful muscle. In a patient with muscular disease, some muscles may be affected by paralysis and not generate force properly. A clinician can use the method to observe muscle function in more detail. Thus, diseased or injured muscles can be detected and treated more accurately. In addition, this proposed method is very useful for designing ergonomic gripping hand tools. To prevent injury under high work-load condition, the design of gripping hand tools should consider load sharing among muscles, taking care not to overuse an extensor muscle group.

In this study, a 3D scanner was used to obtain forearm geometry of each subject. The scanning time took approximately 1 min at a resolution of point-to-point space of approximately 0.6 mm, comprising 250,000 triangles. Using a 3D scanner to construct a forearm model is very practical for clinical application. Body segment geometry data are generally obtained from medical imaging methods such as MRI and gamma-ray scanning (Martin, et al., 1989; Cheng, et al., 2000; Dumas, et al., 2005). However, these methods are expensive and require long measurement times. The advantages of the 3D scanner method over MRI and gamma-ray scanning are low cost and rapid measurement. Comparison of the forearm model constructed from 3D scanner data with the MRI image suggests that the resolution was sufficient to accurately outline the forearm from the geometry obtained. The shapes of the forearm cross-sectional area differed markedly among subjects. The development of a forearm model using a 3D scanner allows comparison with anatomical information.

In the study, we did not include the bone region in the forearm model; however, when our results were compared with MRI of each subject’s forearm, the expected bone area showed very low activity. The prospect of further study to detect the bone region using the EMG–CT method is interesting and important. It was also assumed that force exerted by fingers was equal to total muscle force generated by forearm muscles. However, there are many factors such as wrist and arm posture and the position of applied load on fingers that can affect the muscle force generation mechanism. Gripping may be usable as a calibration process when this method is applied to estimate stress in other hand and finger motions. This method is also limited to static isometric conditions. The relationship between muscle activity and muscle stress was assumed to be linear, an assumption that might not apply under dynamic conditions. Further studies are thus required to establish the effects of dynamic conditions on muscle activity and stress.

In summary, this study has shown high potential for estimating stress distribution generated in the forearm muscles using an EMG–CT band and a 3D scanner. The use of a new EMG conduction model devised from subjects’ forearm geometry and a model for calculating stress from muscle activity represent an improvement in the EMG–CT method and make it more clinically applicable. This improvement allows visualizing stress within the forearm muscles, a capability that may advance the development of diagnostic tools.

References

Buchanan, T. S., Evidence that maximum muscle stress is not a constant: differences in specific tension in elbow flexors and extensors, Medical engineering & physics, Vol.17, No.7 (1995), pp. 529-536.

Buchanan, T. S., Moniz, M. J., Dewald, J. P. and Zev Rymer, W., Estimation of muscle forces about the wrist joint during isometric tasks using an EMG coefficient method, Journal of biomechanics, Vol.26, No.4-5 (1993), pp. 547-560.

Burke, R. E. and Tsairis, P., Anatomy and innervation ratios in motor units of cat gastrocnemius, The Journal of physiology, Vol.234, No.3 (1973), pp. 749-765.

Cheng, C.-K., Chen, H.-H., Chen, C.-S., Lee, C.-L. and Chen, C.-Y., Segment inertial properties of Chinese adults determined from magnetic resonance imaging, Clinical biomechanics, Vol.15, No.8 (2000), pp. 559-566.
Close, R., Dynamic properties of fast and slow skeletal muscles of the rat after nerve cross-union, The Journal of physiology, Vol.204, No.2 (1969), pp. 331-346.

Darling, W. G., Cole, K. J. and Miller, G. F., Coordination of index finger movements, Journal of biomechanics, Vol.27, No.4 (1994), pp. 479-491.

Del Santo, F., Gelli, F., Ginanneschi, F., Popa, T. and Rossi, A., Relation between isometric muscle force and surface EMG in intrinsic hand muscles as function of the arm geometry, Brain research, Vol.1163, (2007), pp. 79-85.

Dennerlein, J. T., Diao, E., Mote, C. D., Jr. and Rempel, D. M., Tensions of the flexor digitorum superficialis are higher than a current model predicts, Journal of biomechanics, Vol.31, No.4 (1998), pp. 295-301.

Disselhorst-Klug, C., Schmitz-Rode, T. and Rau, G., Surface electromyography and muscle force: limits in sEMG-force relationship and new approaches for applications, Clinical biomechanics, Vol.24, No.3 (2009), pp. 225-235.

Dumas, R., Aissaoui, R., Mitton, D., Skalli, W. and de Guise, J. A., Personalized body segment parameters from biplanar low-dose radiography, IEEE transactions on bio-medical engineering, Vol.52, No.10 (2005), pp. 1756-1763.

Duque, J., Masset, D. and Malchaire, J., Evaluation of handgrip force from EMG measurements, Applied ergonomics, Vol.26, No.1 (1995), pp. 61-66.

Hagg, G. M. and Milerad, E., Forearm extensor and flexor muscle exertion during simulated gripping work -- an electromyographic study, Clinical biomechanics, Vol.12, No.1 (1997), pp. 39-43.

Hof, A. L., EMG and muscle force: An introduction, Human Movement Science, Vol.3, No.1-2 (1984), pp. 119-153.

Hoozemans, M. J. and van Dieen, J. H., Prediction of handgrip forces using surface EMG of forearm muscles, Journal of electromyography and kinesiology, Vol.15, No.4 (2005), pp. 358-366.

Johanson, M. E., James, M. A. and Skinner, S. R., Forearm muscle activation during power grip and release, The Journal of hand surgery, Vol.23, No.5 (1998), pp. 938-944.

Johanson, M. E., Skinner, S. R., Lamoreux, L. W., St Helen, R., Moran, S. A. and Ashley, R. K., Phasic relationships of the extrinsic muscles of the normal hand, The Journal of hand surgery, Vol.15, No.4 (1990), pp. 587-594.

Kanda, K. and Hashizume, K., Factors causing difference in force output among motor units in the rat medial gastrocnemius muscle, The Journal of physiology, Vol.448, (1992), pp. 677-695.

Karlsson, S. and Gerdle, B., Mean frequency and signal amplitude of the surface EMG of the quadriceps muscles increase with increasing torque—a study using the continuous wavelet transform, Journal of electromyography and kinesiology, Vol.11, No.2 (2001), pp. 131-140.

Lannergren, J. and Westerblad, H., The temperature dependence of isometric contractions of single, intact fibres dissected from a mouse foot muscle, The Journal of physiology, Vol.390, (1987), pp. 285-293.

Martin, P. E., Mungiole, M., Marzke, M. W. and Longhill, J. M., The use of magnetic resonance imaging for measuring segment inertial properties, Journal of biomechanics, Vol.22, No.4 (1989), pp. 367-376.

Messier, R. H., Duffy, J., Litchman, H. M., Paslay, P. R., Soechting, J. F. and Stewart, P. A., The electromyogram as a measure of tension in the human biceps and triceps muscles, International Journal of Mechanical Sciences, Vol.13, No.7 (1971), pp. 585-598.

Nakajima, Y., Keeratihattayakorn, S., Yoshinari, S. and Tadano, S., An EMG-CT method using multiple surface electrodes in the forearm, Journal of electromyography and kinesiology, Vol.24, No.6 (2014), pp. 875-880.

Pruim, G. J., de Jongh, H. J. and ten Bosch, J. J., Forces acting on the mandible during bilateral static bite at different bite force levels, Journal of biomechanics, Vol.13, No.9 (1980), pp. 755-763.

Schuind, F., Garcia-Elias, M., Cooney, W. P., 3rd and An, K. N., Flexor tendon forces: in vivo measurements, The Journal of hand surgery, Vol.17, No 2 (1992), pp. 291-298.

Stančić, I., Musić, J. and Zanchi, V., Improved structured light 3D scanner with application to anthropometric parameter estimation, Measurement, Vol.46, No.1 (2013), pp. 716-726.

Van den Herrewegen, I., Cuppens, K., Broeckx, M., Barisch-Fritz, B., Vander Sloten, J., Leardini, A. and Peeraer, L., Dynamic 3D scanning as a markerless method to calculate multi-segment foot kinematics during stance phase: methodology and first application, Journal of biomechanics, Vol.47, No.11 (2014), pp. 2531-2539.

Vigouroux, L., Quaine, F., Labarre-Vila, A., Amarantini, D. and Moutet, F., Using EMG data to constrain optimization procedure improves finger tendon tension estimations during static fingertip force production, Journal of biomechanics, Vol.40, No.13 (2007), pp. 2846-2856.