Development of femoral bone fracture model simulating muscular contraction force by pneumatic rubber actuator

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***Abstract*—In femoral fracture reduction, orthopedic surgeons must pull distal bone fragments with great traction force and return them to their correct positions, by referring to 2D-fluoroscopic images. Since this method is physically burdensome, the introduction of robotic assistance is desirable. While such robots have been developed, adequate control methods have not yet been established because of the lack of experimental data. It is difficult to obtain accurate data using cadavers or animals because they are different from the living human body's muscle characteristics and anatomy. Therefore, an experimental model for simulating human femoral characteristics is required. In this research, human muscles are reproduced using a McKibben-type pneumatic rubber actuator (artificial muscle) to develop a model that simulates typical femur muscles using artificial muscles.**

I. INTRODUCTION

In femoral fracture reduction, orthopedic surgeons must pull distal bone fragments with great traction force to return them to their correct positions, by referring to 2D-fluoroscopic images. Since this approach can be burdensome, robotic assistance would be highly desirable. Some robots have been developed for this purpose [1-3] and clinical tests have been performed. [4,5]

However, as yet no control method for such robots has been established because of the following three reasons. First, there are insufficient clinical tests. Second, animal experimentation is ineffective because animals are anatomically different from humans. Third, cadaver experimentation is also ineffective because the mechanical characteristics of human muscular tissues are significantly altered after death.

Control methods are best studied by performing clinical tests; however, there are only few methods to perform these tests. Therefore, an alternative clinical testing method for studying control method is required. One alternative is to use an experimental model that simulates human femoral characteristics. In this paper, we propose an experimental model for studying a control method for orthopedic surgeons to practice femoral fracture reduction. This model imitates the human musculoskeletal system in shape and may be used by many different robots.

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II. PROPOSED SYSTEM AND EXPERIMENT

A. Modeling the Femur Muscles

The femur is almost completely encased in muscles, most of which are attached to the bone itself. The resting muscle tones of the primary muscles attached to and spanning the femur largely determine the observed displacement. Therefore, in our experimental model, these muscle forces on the femur play an important role in performing the anatomical reduction.

Common deformities in a femoral diaphyseal fracture are the shortening, flexion and external rotation of proximal fragments, and the extension of distal fragments. Primary muscles that contribute to fracture displacement include the hamstrings, quadriceps, hip adductors, abductors, external rotators, iliopsoas, and the gastrocnemius muscle. Shortening occurs because of the pull of the hamstrings and quadriceps muscles. The proximal segment is typically flexed, adducted or abducted, and externally rotated by the muscular pull of the iliopsoas, the hip adductors or abductors, and the external rotators, respectively. The distal fragment is typically medialized due to the pull of the adductors. Because of these largely unopposed muscle forces, any attempt to reduce proximal and distal fractures by increasing the distraction force is typically futile. Limb position, strategic bumps, and externally applied forces are much more effective than brute strength in correcting the angulatory and translational deformities that occur. [6]

In this research model, to simplify the problem, the proximal segment is fixed to the pelvis and the parts below the knee are excluded.

The model simulates the representative muscles of the shortening group and the adductors:

- Rectus femoris (RF) (quadriceps muscles, shortening group)
- Biceps femoris long head (BFLH) (hamstrings, shortening group)
- Adductor longus (AL) (adductor muscle)

Furthermore, we also simulate the tensor fasciae latae (TFL) muscle (connected to the iliotibial band), which is antagonistic to the adductors and seems to have significant influence in traction work. The musculoskeletal model for simulation is shown in Fig.1.

B. Mathematical Model of Muscle

Since muscle has nonlinear characteristics, it has many modeling methods, considering its nonlinearity. Hill [7] described a muscle with a model using three elements—

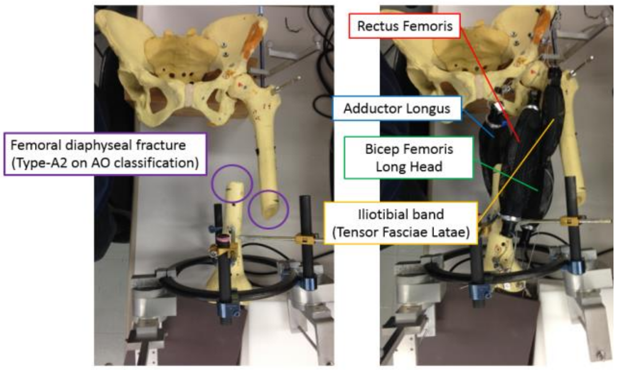


Fig.1 Musculoskeletal model

contractile, serial elastic, and parallel elastic elements. Zajac [8] introduced normalized force and normalized length into Hill's model and developed a more practical model. In this model, muscle force (F) is normalized by peak isometric active force (F_0) and muscle fiber length (L) is normalized by an optimal fiber length (L_0), which is the length when the muscle generates peak isometric active force. Fig.2 shows normalized force–length muscle properties.

In this research, we use the force–length properties, shown in Fig.2, in our mathematical model and in the parameters of muscle used in OpenSim, the open source musculoskeletal simulator developed by Delp et al. [9] In addition, we introduce activation coefficient "a," which expresses the activity state of a muscle with numbers from 0 to 1 and controls the active force by being multiplied with it. To simplify the problem, we do not consider the expansion and contraction of tendons.

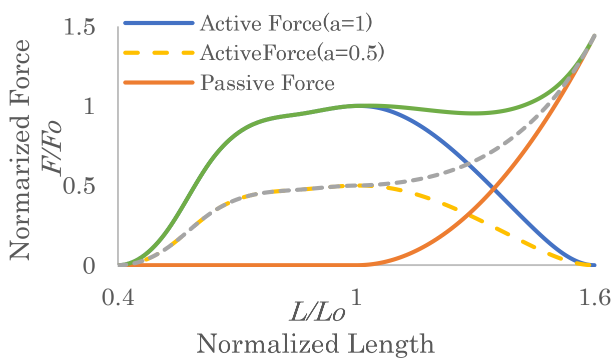


Fig.2 Force-length properties

C. McKibben-Type Artificial Muscle

Because of its similarity with muscle characteristics, we chose the McKibben-type artificial muscle as the actuator for our simulation model. The McKibben-type artificial muscle is a kind of pneumatic rubber actuator made of thin rubber tubing covered by a braided mesh sleeve and clamped at both ends by a metal ring. When supplied with compressed air, the artificial muscle expands and contracts in the radial and axial directions, respectively, thereby producing traction forces. The reason for selecting the McKibben-type artificial muscle is that it has many advantages, such as a high force–weight ratio, simple and flexible structure, and good compliance.

When the supply of air pressure is constant, our model responds like the Hill's muscle model with respect to the force–velocity relationship for muscle contraction, but differs from Hill's muscle model in its force–length properties. [10] Therefore, to simulate the force–length properties of Hill's muscle model, we must carefully control the supply of air pressure to the artificial muscle.

The steps for simulating muscle are as follows:

1. Calculate the force–length-air pressure relationships of the McKibben artificial muscle by the isometric contraction test.
2. Make a force–length curve from the Hill's model and the existing muscle parameters.
3. Control the supply of compressed air to accurately reproduce the force–length curve of the muscle model.

In this research, force–velocity properties are not considered because muscle length does not change drastically in fracture reduction. Moreover, we use open-loop force control by length rather than closed-loop force control.

D. System Properties

Fig.3 provides an overview of the experiment. The femur musculoskeletal model (Fig.1) is fixed to a table. Optical tracking markers (NDI Polaris) are fixed to both the proximal and distal bone fragments. A carbon ring with handles for fracture reduction is also fixed to the proximal bone fragment. When this handle is pulled for fracture reduction, the force and torque can be measured by a 6-axis force sensor (Leptrino CF055CA501U) attached to the handle. The position of both ends of the artificial muscles is measured by each of the tracking markers. We measured the marker positions at each end in local coordinates before the test. The global coordinates of the marker positions at each end are calculated by coordinate transformation using the position and attitude of the markers measured by the optical tracking sensor during the experiment. We then calculated muscle length by the positions of both muscle ends and determined the amount of compressed air to be supplied to the muscle.

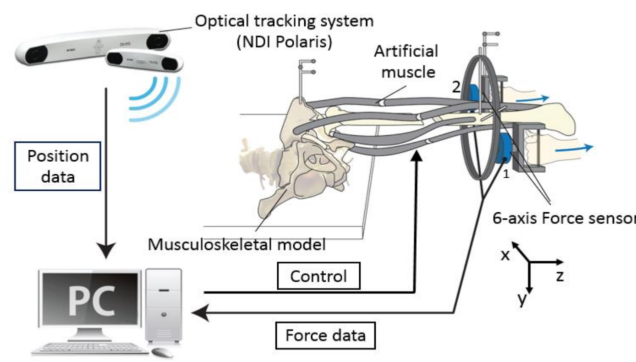


Fig.3 Experiment overview

E. Experiment

We performed two experiments.

1. Contraction test of a single artificial muscle

The purpose of this experiment was to evaluate the control method. First, we performed an isometric contraction test and acquired the force–length-air pressure relationship. Then, we controlled the artificial muscle and compared the measured force, the desired force, and the force at constant air pressure.

2. Femoral fracture reduction experiment

We performed fracture reduction in experimental model and measured the force while the distal bone fragment was being pulled. We performed this experiment under controlled and uncontrolled (constant air pressure) conditions and finally compared the results.

III. RESULT

A. *Contraction Test of a Single Artificial Muscle*

Fig.4 shows the results of the artificial muscle isometric contraction test (to simulate RF, AL, and BFLH; TFL is simulated by another artificial muscle) with a tension testing machine (Shimadzu EZ-SX).

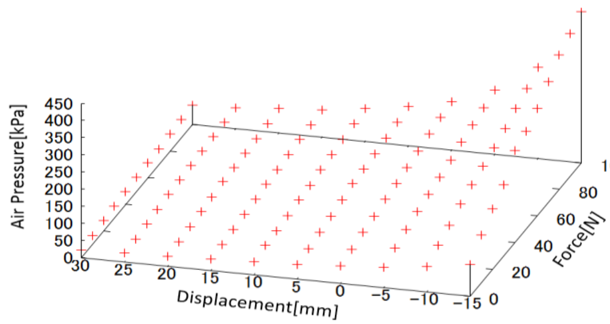


Fig.4 Force-length-air pressure relationship
(Displacement is based on the natural length of artificial muscle)

We performed a contraction test of a single controlled artificial muscle using this result (Shimadzu EZ-SX, expand speed: 0.5 mm/s).

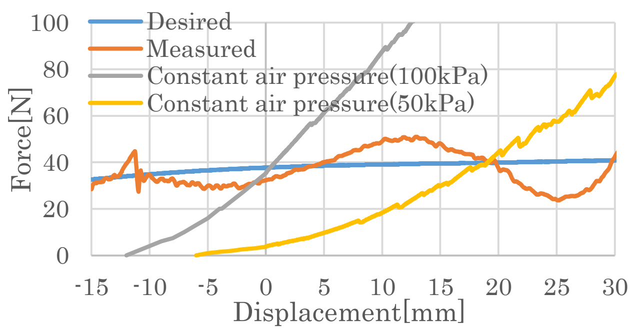
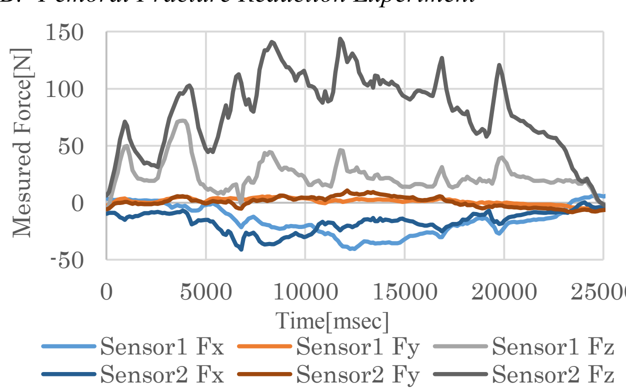


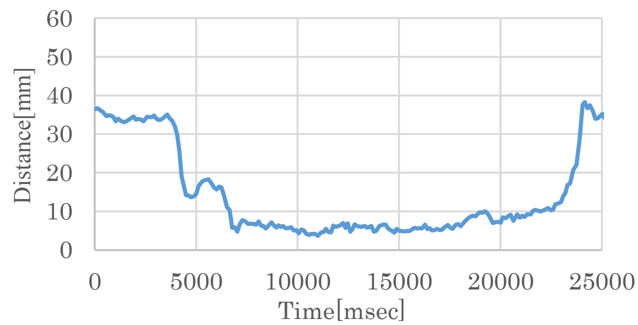
Fig. 5 Contraction test of a single artificial muscle

In Fig.5, an error between the desired force and the measured force can be observed. The reason for this error was that the force–length-air pressure relationship was measured under isometric contraction, and there was an approximation error in the isometric contraction test.

B. *Femoral Fracture Reduction Experiment*



(a)



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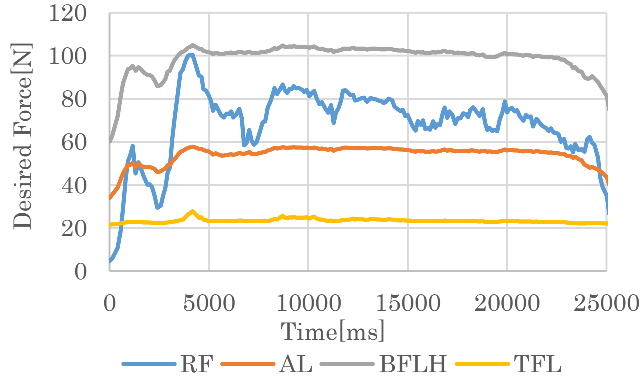
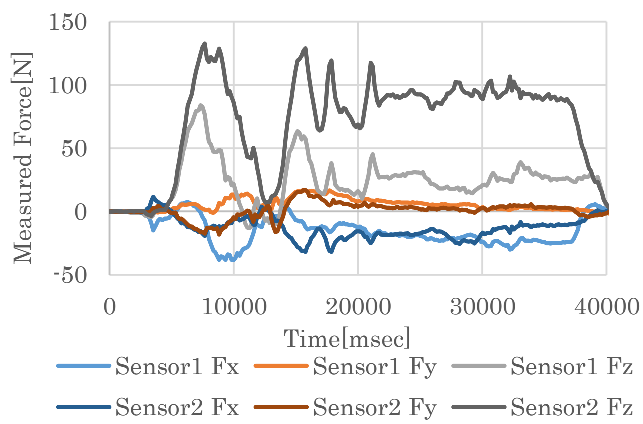
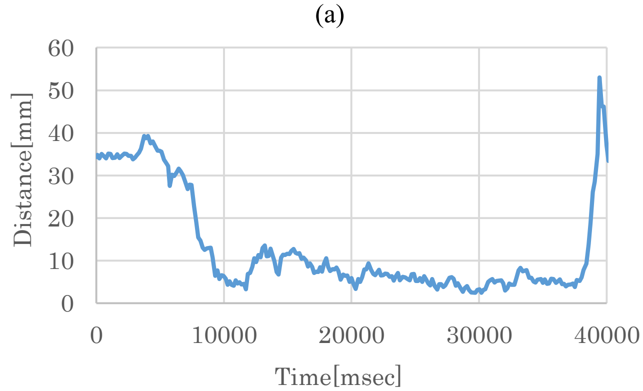


(c)

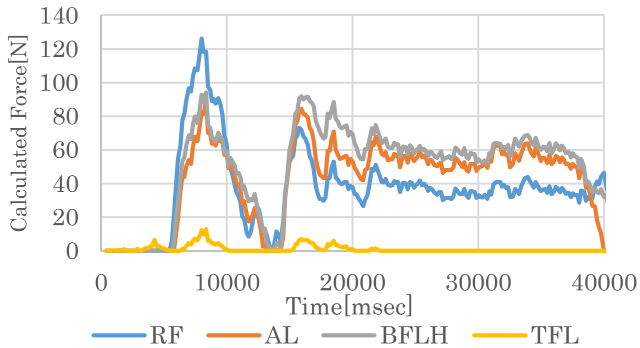
Fig.6 Experiment result at a = 0.15
(a) Time series data of force measured by force sensors (b) Time series data of distance ("Distance" denotes the distance between the proximal and distal bone fragments, does not denotes muscle length) (c) Time series data of desired force



(a)



(b)



(c)

Fig.7 Experiment result at constant air pressure (80kPa)
(a) Time series data of force measured by force sensors (b)Time series data of distance (c) Time series data of calculated force (Calculated force is calculated from muscle length and air pressure)

In Fig.6 and Figs.7, the experimental results at a=0.15 and constant air pressure (80kPa) are shown. Comparing Figs.6 and 7, the measured forces are not significantly different (Fig.6(a), Fig.7(a)); however, the estimated muscle forces exhibit great differences (Fig.6(c), Fig.7(c)).

IV. Discussion

In this research, our model used open-loop air supply control by the length of the artificial muscles, as measured by the optical tracking sensor. Thus, according to Fig.6, the air supply was not strictly controlled. However, our results were more similar to the mathematical model of a muscle's mechanical characteristics than in uncontrolled models, where air pressure is held constant. Furthermore, this error in our model can be removed using closed-loop force control to introduce a force sensor to the artificial muscle. Therefore, in future experiments, we will remove this error using closed-loop force control if the influence of the error due to open-loop control is found to be not negligible.

Furthermore, we have used an optical tracking sensor, which has some advantages. For example, there is no interference during the experiment, and the position and attitude of the bone fragments and length of each artificial muscle can be measured with only two tracking markers. On the other hand, the disadvantages of using this sensor are that in some positions or attitudes, the tracking marker cannot be detected. If the object to which the tracking marker is fixed becomes distorted, the sensor will not work properly because of the change in local position. These problems can be solved by increasing the number of tracking sensors and tracking markers and performing simultaneous measurements. However, because optical tracking sensors are expensive, an alternative method is required for general use.

The one orthopedic surgeon who tested our system found that it had a similarity and a difference to the actual human body. The elasticity was similar to that in real fracture reduction in the cases of a = 0.1 to a = 0.15. However, at the beginning of traction, the doctor felt that the artificial muscle was softer than that in a real fracture reduction in all cases. This difference is caused by the small range of motion of the actuator. Although the actual human muscle (RF) stretches from −110 mm to 50 mm based on optimal muscle length, the artificial muscle in this study operates at −15 mm to 30 mm based on its natural length. Therefore, our system should be used for simulation around the fracture site only.

V. Conclusion

We developed a femoral fracture reduction model that simulates human femoral characteristics using the McKibben-type artificial muscle and confirmed its working. An orthopedic surgeon also evaluated this model and provided valuable information about its performance. In future work, we plan to investigate the influence of the error from open-loop control of the air supply and to modify the model and control method with reference to surgeon's evaluation. In addition, we will apply this model to other fracture types. We conclude that the mathematical model simulated on the computer can be implemented and its results will be compared with clinical data to improve the model toward the establishment of an effective robotic control method for fracture reduction.

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