ACTIVE MUSCLE FORCE AND MOMENTS IN MAJOR HUMAN JOINTS

by

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INTRODUCTION

It is a well established fact that mathematical models play a very significant role in predicting the biodynamic response of the human body subjected to vehicular crash conditions. Properly developed models of the human body provide a sound basis for the design of support-restraint systems and vehicles as well. These mathematical models recently have attracted particular attention in vehicle crash victim studies in view of the high cost of the experiments with cadavers or anthropometric dummies. The most sophisticated versions of the total-human-body models are articulated and multi-segmented to simulate all the major joints and segments of the human body in three-dimensional space.

During the last decade and a half there has been as many as ten distinct vehicle-occupant models developed in the U.S.A. alone. Formulation of the equations of motion in these models has been done by utilization of Lagrange's equations, Euler's rigid body equations, and Lagrange's form of d'Alembert's principle. Naturally, the first models developed have been two-dimensional and they take their impetus from the original work of McHenry (1). Since this work, refinements and other two-dimensional models appeared in the crash victim simulation literature (2,3,4,5,6). The three-dimensional models developed in various research centers in the USA include three-segment model of HSRI (7) in Michigan, twelve-segment models of TTI (8) in Texas and of UC (9) in Ohio, and fifteen-segment model of Calspan (10,11,12) in New York. The Calspan three-dimensional model in recent years went through several development stages. Its present version can be considered the most sophisticated three-dimensional crash victim simulation model available at the present time. It can have any desired number of body segments limited only by the storage capability of the computer. Multiple impacts between the crash victim and vehicle, modeling of pedestrian struck by vehicle, as well as modeling of multiple occupants of a vehicle are some of the salient features of the Calspan model. By some additional features the Calspan model is now also available for air force related applications. An excellent review of both two- and three-dimensional mathematical models simulating biodynamic response of the human body was provided by King and Chou (13).

Short-time response of the multi-segmented models to predict accurately live human response requires proper characterization of the passive resistive force and torque data in major articulating joints. A research program developed to collect both passive and active resistive force and torque data on major articulating joints has already been described (14). Results on passive resistive torques associated with the rotational motion of body segments about long bone axes of the upper and lower extremities (15), and results on the active muscle torques about the same long-bone axes (16) have been provided. Simulation of biodynamic events lasting more than a fraction of a second also requires the incorporation of active muscles into the multi-segmented models and constitute long-time response of the model. The next generation models of the human body will most likely have, in addition to the passive resistive soft-tissue elements, contributions of various active muscles in determining the motion of one body segment with respect to the adjacent body segment. This paper is concerned with the collection of active muscle force data for the upper and the lower limbs. Some representative results are presented from experiments conducted on 20-22 year-old three male and three female subjects to determine their isometric muscle resistance at different orientations of the limbs with respect to the torso.

MATERIALS AND METHODS

In this research, both the position determination of the limbs and force data collection were accomplished by utilizing several sonic emitters and a sonic digitizer. This new technique is quite different than the one used in previously reported studies (14,15,16) where the kinematic and force data were obtained by means of three-dimensional linkage devices. The major components of the experimental apparatus for this research are a subject restraint system, a force application device which employs three sonic emitters, and an upper arm or upper leg cuff with four sonic emitters. Figure 1 shows the microphone assembly, the sonic emitters, and the rest of the experimental setup. The sonic emitters are utilized to determine the direction and the location of the force application on the limbs and the orientation of the upper arm and the upper leg with respect to the torso. Since principles of sonic digitizing, description of a force and moment transducer and its usages with a force applicator having sonic emitters, and position determination by means of sonic emitters have already been presented in a previous article (17), they are not repeated here.

Subjects were university students with no special training in athletics. Selected anthropometric measurements of the subjects along with definitions of the anthropometric measurements are given in (17). Two sets of experiments on the collection of active muscle force and moment data were considered. In the first set of experiments, isometric muscle force resistances and corresponding moment values at the shoulder joints of subjects were determined at different positions of the upper arm with fully extended elbow. Various orientations of arm were described by means of the spherical coordinates θ and ϕ. The θ angle refers to the angle between the z-axis of the torso and the long-bone axis of the upper arm. This angle also defines the shoulder flexion-extension in the sagittal plane. The ϕ angle refers to the angle between the projection of the long-bone axis of the upper arm on the xy-plane and the x-axis; the positive and negative values of this angle also define the shoulder abduction and adduction, respectively. Note that in Fig. 1 the torso fixed Cartesian coordinate system is shown as (x_t, y_t, z_t) for which positive x-direction refers to anterior direction, positive y-direction to right lateral and, naturally, positive z-direction refers to inferior direction. The angles θ_1, ϕ_1 and θ_2, ϕ_2 shown in Fig. 1 define the angular orientation of the subject restraint system and the microphone assembly, respectively, with respect to the laboratory coordinate system (X,Y,Z).

In the second set of experiments the isometric muscle force resistances and corresponding moment values at the hip joints of subjects were de-

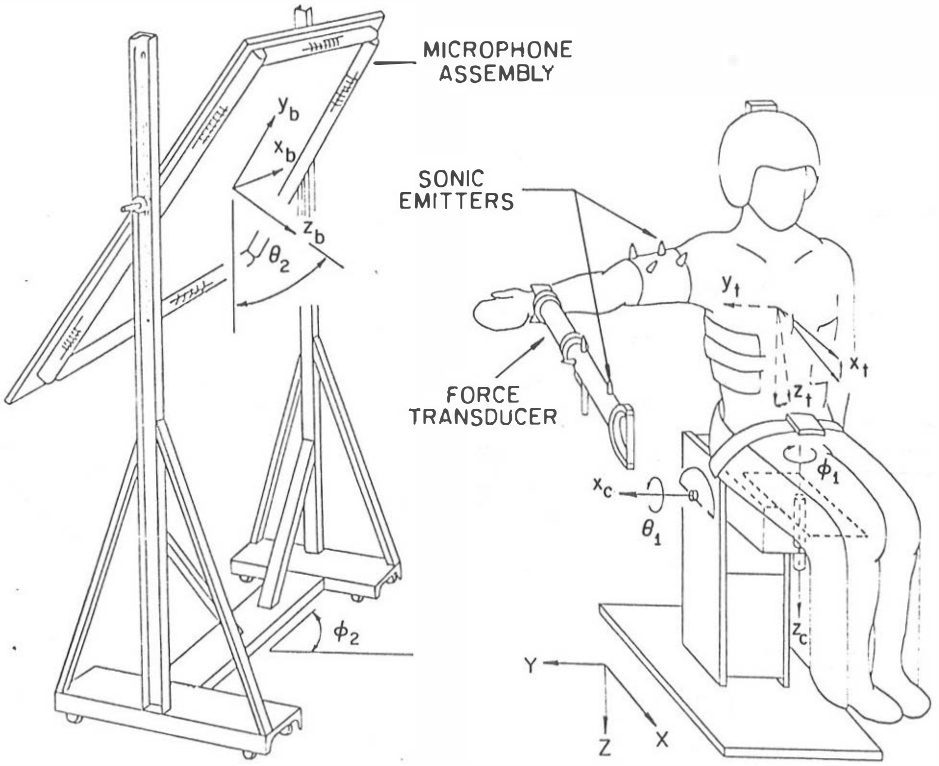

Fig. 1 Schematic drawing of experimental setup.

termined at different positions of the lower limbs. Tests were conducted for various combinations of the spherical coordinates θ and ϕ, describing the position of the upper leg in relation to the torso with the knee flexed and the knee locked positions. The θ angle refers to the angle between the z-axis of the torso and the femoral axis ($\theta = 90°$ for seated position; $\theta = 0°$ for standing position). The angle ϕ measures the projection of the femoral axis on the xy plane and the x-axis of a coordinate system fixed in the mid-torso location. The isometric force applications by the subjects were on the wrist and elbow for the first set and on the knee and just above the ankle for the second set. The force application by the subjects lasted approximately three seconds and tests were repeated twice for each subject.

RESULTS AND CONCLUDING REMARKS

In this paper, because of the space limitations, only the maximum values of the isometric resistive muscle forces of the subjects tested are presented. The corresponding moments about the shoulder and the hip joints can be found in the final report of a recently completed research project (18). It should be emphasized that the numerical values plotted in Figs. 2-6 are the magnitudes of the force vectors having all three components although, depending upon the direction of the force application, one component of the force vector predominates the other two.

Based on the numerical results presented in Figs. 2-6, several concluding remarks can be made on the maximum values of the isometric resistive muscle force response of the upper and lower limbs of the human body:

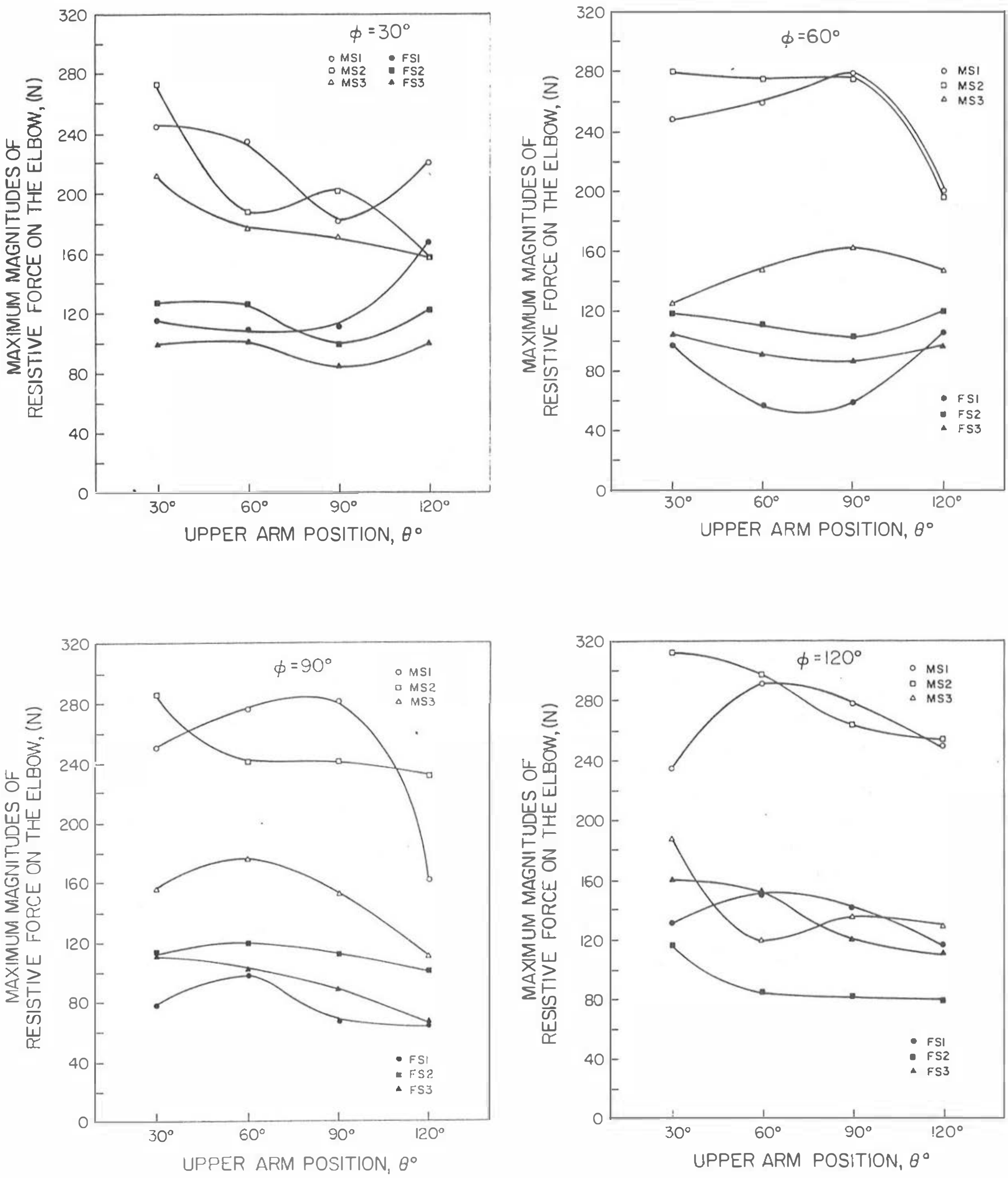

Fig. 2 Maximum values of the isometric resistive muscle force by the male (MS1, MS2, MS3) and the female (FS1, FS2, FS3) subjects against the external force applications on the elbow at various arm positions in $\phi = 30°$, $60°$, $90°$, and $120°$ planes.

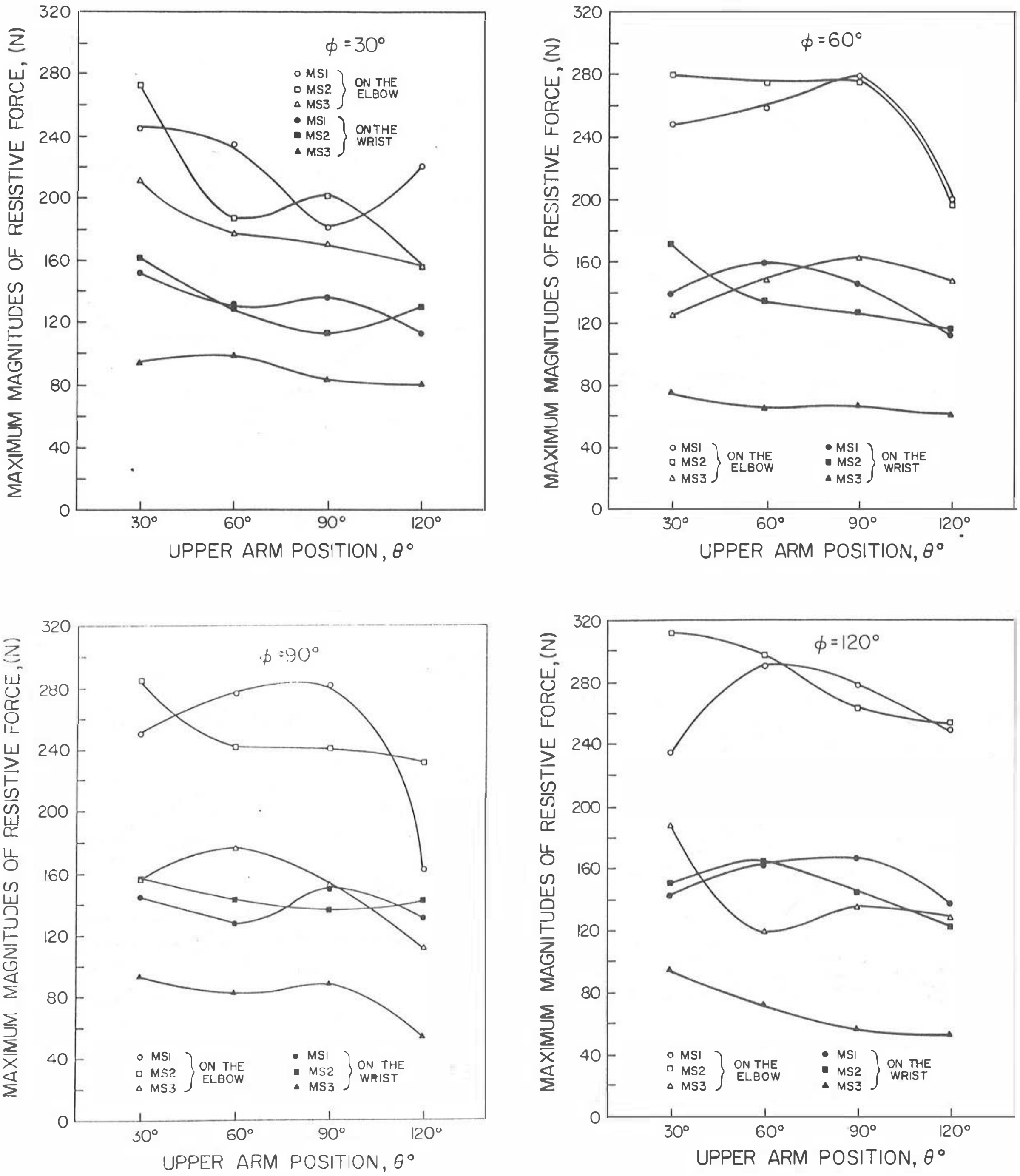

Fig. 3 Maximum values of the isometric resistive muscle force by the male subjects against the external force applications on the elbow and the wrist at various arm positions in ϕ = 30°, 60°, 90°, and 120° planes.

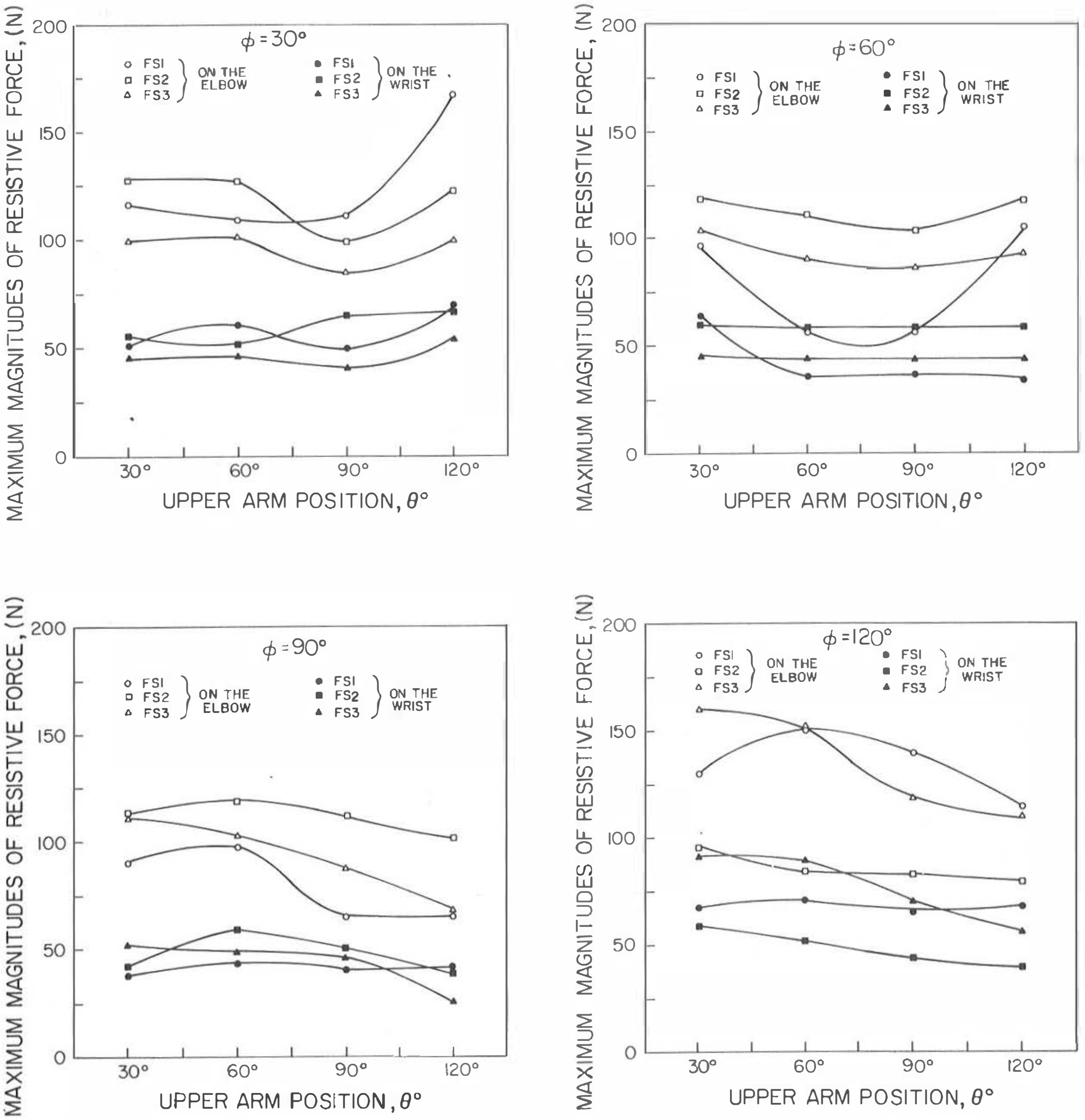

Fig. 4 Maximum values of the isometric resistive muscle force by the female subjects against the external force applications on the elbow and the wrist at various arm positions in ϕ = 30°, 60°, 90°, and 120° planes.

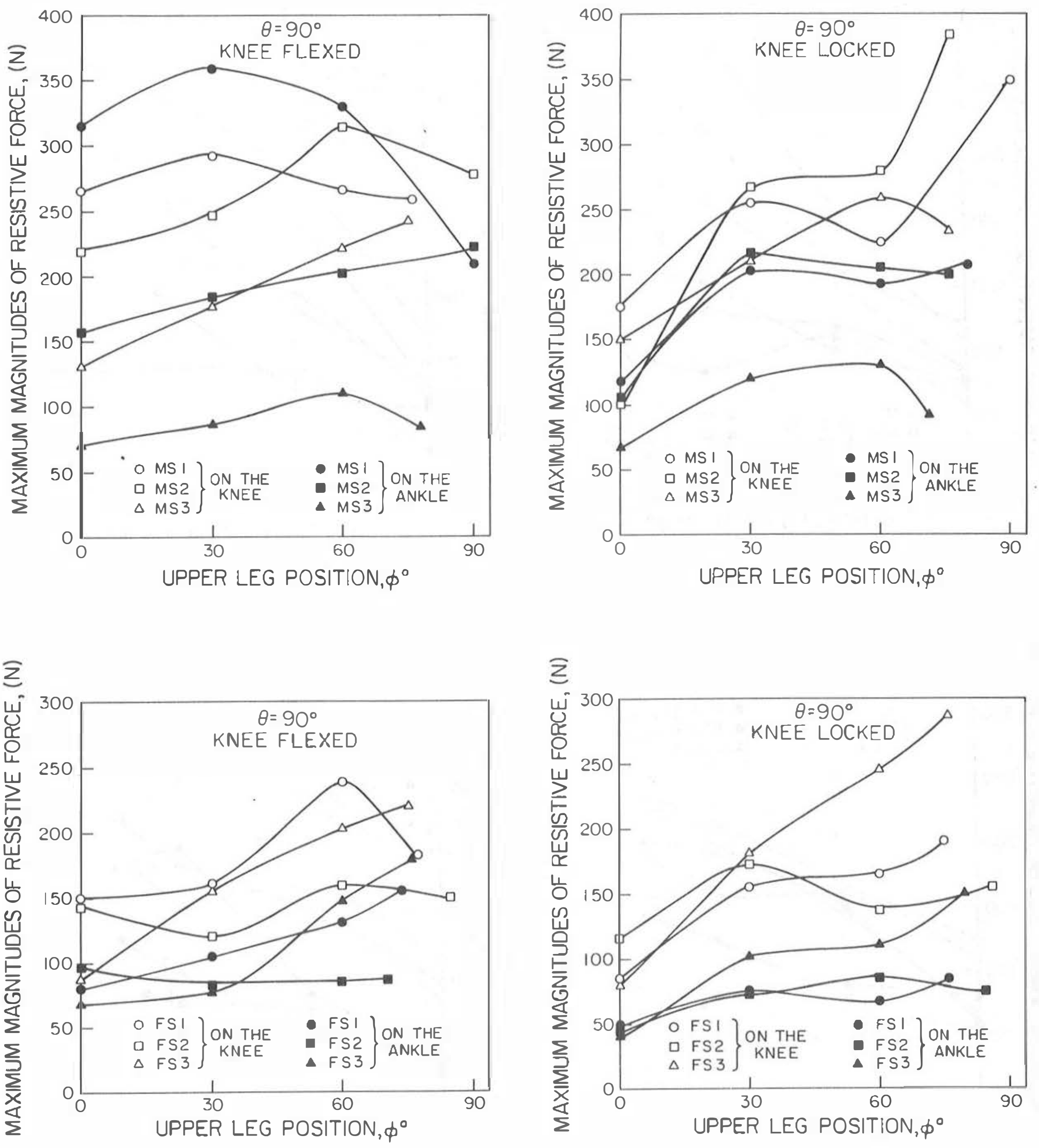

Fig. 5 Maximum values of the isometric resistive muscle force by the male (MS1, MS2, MS3) and the female (FS1, FS2, FS3) subjects for the knee flexed and the knee locked configurations for various lateral positions of the upper leg at θ = 90°.

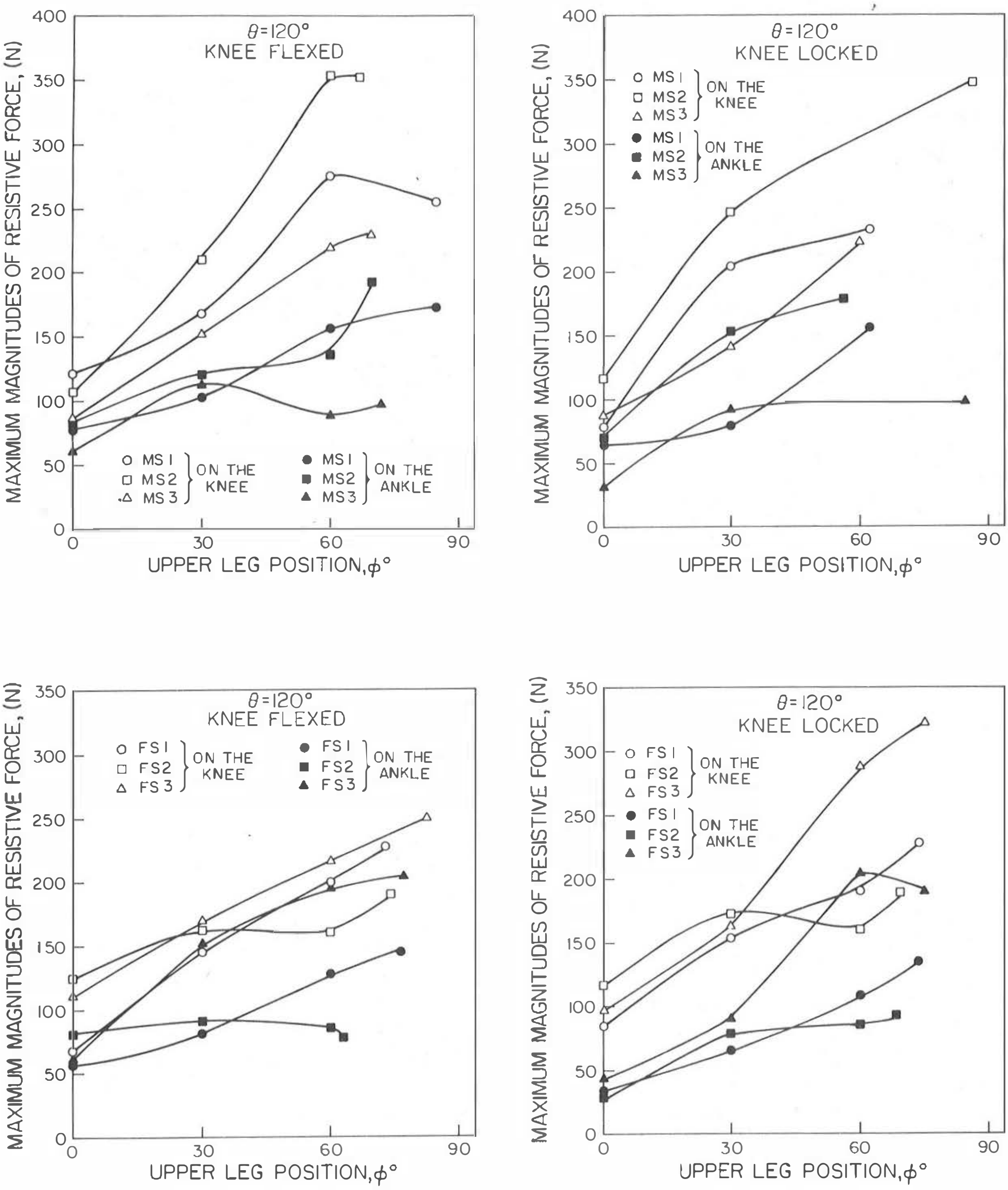

Fig. 6 Maximum values of the isometric resistive muscle force by the male (MS1, MS2, MS3) and the female (FS1, FS2, FS3) subjects for the knee flexed and the knee locked configurations for various lateral positions of the upper leg at $\theta = 120°$.

to increase muscular strength. It is reasonable to expect that the values of the isometric resistive muscle forces of the upper and lower limbs depend not only on sex and anthropometry of the subjects as shown in this research but also on age, physical fitness, and, particularly, degree of training for muscular strength. We can also point out that, although there are both intra- and inter-subject variations for the maximum values of the resistive muscle forces of the extremities, there are some trends one can establish for the behavior of their magnitudes.

It is expected that incorporation of the active muscle force data into the multi-segmented mathematical models of the human body should improve the long-time response capabilities of these models so that they can simulate more realistically the biodynamic events which take place when the human body is subjected to vehicular crash conditions. Improvements of the mathematical models with incorporation of the active muscle force data results from the notion that calculated deceleration profiles of the limbs with respect to the torso will be substantially different than those calculated from a model including only the passive resistive force and moments in major articulating joints. Recognizing that dislocations of the major articulating joints along with associated soft and hard tissue injuries are closely related to the magnitudes of the deceleration forces, one can appreciate the importance of incorporating active muscle force and corresponding moment data into the multi-segmented mathematical models of the human body.

ACKNOWLEDGEMENTS

This work was supported by the U.S. Air Force Office of Scientific Research and a NATO Grant (No. 012.80).

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1. The average values of the maximum magnitudes of the isometric resistive forces, against the external force applications on the elbow, for the various positions of the arm with a fixed value of $\phi = 30°$, start with 213 N at $\theta = 30°$ and decrease steadily to 179 N at $\theta = 120°$ for the male subjects, for the female subjects, the same quantity starts with 115 N at $\theta = 30°$, exhibits a gradual decrease to 98 N at $\theta = 90°$, and subsequently increase to 146 N at $\theta = 120°$. The ratios of the average values of the maximum magnitudes of the isometric resistive forces against the external force applications on the elbow to that of the wrist range from 1.58 to 1.68 for the male subjects, and 1.96 to 2.33 for the female subjects.

2. The average values of the maximum magnitudes of the isometric resistive forces, against the external force applications on the elbow for the various positions of the arm with a fixed value of $\phi = 60°$, start with 218 N at $\theta = 30°$ and shows a gradual increase to 240 N at $\theta = 90°$ and subsequent drop to 182 N at $\theta = 120°$ for the male subjects; for the female subjects, the same quantity starts with 107 N at $\theta = 30°$, decreases gradually to 82 N at $\theta = 90°$ and shows an increase to 106 N at $\theta = 120°$. The ratios of the average values of the maximum magnitudes of the isometric resistive forces against the external force applications on the elbow to that of the wrist range from 1.71 to 2.12 for the male subjects, and 1.76 to 2.39 for the female subjects.

3. The behavior of the average values of the maximum magnitudes of the isometric resistive forces, against the external force applications on the elbow, for the various positions of the arm with a fixed value of $\phi = 90°$ (in the frontal plane) is very similar for both male and female subjects. This quantity starts with 231 N for the male subjects (101 N for the female subjects) at $\theta = 30°$ and after a slight increase at $\theta = 60°$, it decreases gradually to 169 N (77 N for the female subjects) at $\theta = 120°$. The corresponding elbow to wrist resistive force ratios range from 1.56 to 1.98 for the male subjects, and 1.85 to 2.26 for the female subjects.

4. The behavior of the average values of the maximum magnitudes of the isometric resistive forces, against the external force applications on the elbow, for the various positions of the arm with a fixed value of $\phi = 120°$ is very similar for both male and female subjects. This quantity starts with 245 N for the male subjects (129 N for the female subjects) at $\theta = 30°$ and gradually decreases to 210 N (101 N for the female subjects) at $\theta = 120°$. The corresponding elbow to wrist resistive force ratios range from 1.77 to 2.01 for the male subjects, and 1.77 to 1.96 for the female subjects.

5. For the initial seated position ($\theta = 90°$, $\phi = 0°$), the maximum magnitude of the medially applied isometric resistive force by the leg is greater when the knee is flexed than when the knee is locked, greater for force applied to the knee than for force applied to the ankle, and greater for the male subjects than for the female subjects. In terms of ratios, we find that the ratio of the average values of the maximum magnitudes of the isometric resistive forces with the knee flexed to the force with the knee locked is 1.61 for the male subjects, and 1.67 for the female subjects, when the force application is on the ankle. When the force is applied at the knee, the isometric resistive force of the leg is 1.47 times the force when applied on the ankle for the male subjects, and for the female subjects, this ratio is 1.93, where, for both male and female subjects, the knee is locked. When the knee is flexed, the ratios of force on the knee to force on the ankle are 1.24 for males, and 1.55 for females.

The results presented in this paper were obtained from three male and three female subjects who are young, healthy and with no special training