ASSISTED NECK MOVEMENT UNDER HIGH G+ SITUATIONS
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Abstract
The objective of this research is to investigate smart actuator technology for an adaptive neck brace to assist pilots during high g+ maneuvers. Analysis and experiments were conducted to develop a simplified analytical model of the neck, determine the effectiveness of using a brace to prevent neck injury under high acceleration loads, identify actuator technology capable of actuating the mechanism effectively and efficiently, and to experimentally validate the chosen mechanism and actuator. The experiment successfully demonstrated that an adaptive neck brace can successfully reduce the displacement experienced by acceleration forces. The physical model utilized a scissor mechanism brace configuration using shape memory alloy (SMA) wire for actuation. The presence of the brace avoided high resonant peaks and introduced damping to the original model.

Introduction and Background
The existence of neck injury for fighter pilots has become well documented in the past decade. It is widely accepted that these injuries are primarily a result of the intense acceleration forces experienced by a pilot during a typical sortie. Methods of study undergone to quantify sustained G force effects range to include monitoring the electromyography (EMG) of pilots during these extreme conditions, questioning the pilots on perceived discomfort in flight, comparing neck muscle strength between pilots and non-pilots, as well as investigating medical procedures undergone by pilots recovering from G force induced injuries. Typical injuries resulting from G force induced muscle strain vary from mild sprains to flight career ending debilitating conditions, and have been shown to be experienced by 90% of fighter pilots during their career. Research has concluded that these injuries are more likely to occur when the head is away from the neutral position (the head rested squarely on the shoulders looking forward), as this greatly increases the loads directly supported by the neck. This danger drastically increases once the neck rotates past a 35 degree angle. At such angles muscle activation has been measured as high as 257% maximum voluntary isometric contraction. One study found the pilots head to be away from the neutral position 68% of the time during air combat, quantifying the large amount of time fighter pilots endure these dangerous neck loading conditions. As a result of fatigue, fighter pilots will often unload imposed G forces before rotating their neck, often compromising the effectiveness with which they can achieve their mission or carry out evasive maneuvering.

As fighter aircraft becomes increasingly maneuverable and technology continues to add to the weight of the pilot’s helmet (i.e. Head-Mounted displays), the problem of cervical spine injury promises to become more prevalent.

Despite the vast amount of research dedicated to studying the affect of G forces on pilots, little work has been done to identify preventative devices and mobility aides to assist pilots under these extreme conditions. Most of the methods studied to combat injury involve undergoing workout regimes to increase the strength and flexibility of the muscles supporting the neck. One study attempted to quantify the benefits of using a lumbar support around the lower back. The results revealed a slight benefit in comfort for some pilots while causing discomfort for others. For the pilots that experienced increased comfort, improvement in EMG readings proved too small to be considered statistically relevant.

Objectives
As seen from above, the high maneuverability of modern aircraft poses the constant threat of neck injury to fighter pilots. As aircraft performance continues to improve, this threat is becoming even

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more imminent. Therefore, the objective of the proposed research is to identify potential technologies to combat the threat of neck injury and ease head mobility for pilots under high sustained G situations. More specifically, this research will (1) develop a simplified analytical model of the neck using rigid body mechanics, (2) identify a potential brace mechanism capable of dynamically changing its stiffness, (3) identify actuator technology that can meet the required performance objectives (e.g. speed, displacement, power, simplicity) and (4) experimentally validate the identified system using a simple testbed.

**Methods**

**Mathematical Model**

There are several methods for analyzing the cervical spine, including in vitro, lumped parameter, and finite element analysis. In vitro analysis consists of studying the characteristics of actual spinal columns through controlled tests, advanced imaging, or cadaver dissection and allows for the investigation of the biomechanical behaviors of the cervical spine tissues and geometries.10 Lumped parameter models ignore vertebrae geometry and tissue formation and use simple spherical shapes to represent the vertebrae.10 Finally, finite element models are digital models of the spine that allow for the study of multiple geometries and material properties.10 A finite element model adapted from Deng and Goldsmith will be used in this study for the analysis.

The finite element model used for this study includes all seven cervical vertebrae, C1-C7, and the first two vertebrae of the thoracic region, T1 and T2. The C1 vertebrae and skull are considered as one mass for model simplification. The implemented coordinate system assumes the yz-axis to point vertically upward, the z-axis to point horizontally toward the front of the vertebrae, and the x-axis out the right side of the vertebrae. The axes originate in the center of the vertebrae.

The equations of motion adapted from Deng and Goldsmith (Equation 1) were developed for the lowest three vertebrae, with the bottom vertebrae, T2, considered fixed. Once the pattern created from these three vertebrae could be recognized, it was then extrapolated to develop the equations of motion for all nine vertebrae.

\[
\begin{bmatrix}
 m \ddot{x} \\
 m \ddot{y} \\
 m \ddot{z} \\
 L_{xx} \ddot{\theta}_x \\
 L_{yy} \ddot{\theta}_y \\
 L_{zz} \ddot{\theta}_z
\end{bmatrix} =
\begin{bmatrix}
 k_{11} & 0 & 0 & 0 & k_{15} & k_{16} \\
 0 & k_{22} & k_{23} & k_{24} & 0 & 0 \\
 0 & k_{32} & k_{33} & k_{34} & 0 & 0 \\
 0 & k_{42} & k_{43} & k_{44} & 0 & 0 \\
 k_{15} & 0 & 0 & 0 & k_{55} & k_{56} \\
 k_{16} & 0 & 0 & 0 & k_{65} & k_{66}
\end{bmatrix} \begin{bmatrix}
 x \\
 y \\
 z \\
 \theta_x \\
 \theta_y \\
 \theta_z
\end{bmatrix}
\]

In the above equations, \( k_{ij} \) represents the stiffness coefficient, \( x, y, \) and \( z \) the translational displacements, \( \theta \) the rotational displacement, \( m \) the mass, and \( I \) the moment of inertia.

Using values from Table 1 in the Appendix, also adapted from Deng and Goldsmith, the mass matrix was created by placing mass and moment of inertia values in the diagonals of the matrix. The values were placed to correspond with the vertebrae degrees of freedom they act upon. An example, showing the first six degrees of freedom of this mass matrix, is displayed in Equation 2. The values on the table represent the cervical vertebrae masses, moments of inertia, local coordinate origins, mass center coordinates, and vertebrae orientation respectively.

\[
\begin{bmatrix}
 m_1 & 0 & 0 & 0 & 0 & 0 \\
 0 & m_1 & 0 & 0 & 0 & 0 \\
 0 & 0 & m_1 & 0 & 0 & 0 \\
 0 & 0 & 0 & I_{x1} & 0 & 0 \\
 0 & 0 & 0 & 0 & I_{y1} & 0 \\
 0 & 0 & 0 & 0 & 0 & I_{z1}
\end{bmatrix} \begin{bmatrix}
 x_1 \\
 y_1 \\
 z_1 \\
 \theta_{x1} \\
 \theta_{y1} \\
 \theta_{z1}
\end{bmatrix}
\]

The stiffness coefficients provided in Deng and Goldsmith for compressive loading of the joint between the C2 and C3 vertebrae are displayed in Equation 3.

\[
[k'] = \begin{bmatrix}
122 & 0 & 0 & 0 & 31 & 82 \\
0 & 1083 & -10 & -365 & 0 & 0 \\
0 & -10 & 50 & -78 & 0 & 0 \\
0 & -36.5 & -78 & 1856 & 0 & 0 \\
31 & 0 & 0 & 0 & 1490 & 10 \\
82 & 0 & 0 & 0 & 10 & 1719
\end{bmatrix}
\]

It is possible to derive the stiffness coefficients for the other vertebrae proportionately based upon their joint cross sectional areas, but due to these proportions varying greatly between individuals and this study’s need for only a representative model, these cross sectional areas were considered equal.
Using the stiffness matrix determined from the process above, a state space analysis was developed using MATLAB. The A and B matrices were defined as displayed in Equations 4 and 5, where \([M]\) represents the mass matrix, \([K]\) represents the stiffness matrix, \([C_g]\) represents the damping matrix, and \([F]\) represents the force vector.

\[
A = \begin{bmatrix}
0 & 1 \\
[M]^{-1}[K] & [M]^{-1}[C_g]
\end{bmatrix}
\] (4)

\[
B = \begin{bmatrix}
0 \\
[M]^{-1}\{F\}
\end{bmatrix}
\] (5)

In this model the damping matrix, \([C_g]\), was created using Equation 6, where \(\alpha\) and \(\beta\) are constants adjusted to best represent the Deng and Goldsmith model.

\[
[C_g] = \alpha[M] + \beta[K]
\] (6)

The state space model was used to determine the effects of increasing neck stiffness upon an applied load. It was determined that a brace stiffened in the \(z\) and \(\theta_x\) directions would be the most beneficial to the pilot, as it would minimize the pilots needed effort to stay in the upright position. To achieve neck stiffening in only these directions, the terms represented by a “1” in Equation 7 were increased by a multiple of the original number, correlating to a stiffening of the \(z\) and \(\theta_x\) terms throughout the complete stiffness matrix.

\[
\begin{bmatrix}
0 & 0 & 1 & 1 & 0 & 0 \\
0 & 0 & 1 & 1 & 0 & 0 \\
1 & 1 & 1 & 1 & 1 & 1 \\
1 & 1 & 1 & 1 & 1 & 1 \\
0 & 0 & 1 & 1 & 0 & 0 \\
0 & 0 & 1 & 1 & 0 & 0
\end{bmatrix} \times \begin{bmatrix}
x \\
y \\
z \\
\theta_x \\
\theta_y \\
\theta_z
\end{bmatrix}
\] (7)

Next, acceleration forces were imposed on each vertebrae in the \(-y\) and \(-z\) directions, representing the forces a pilot would need to overcome to raise out of a seat back position. The force was imposed using an identical triangular input response to the Deng and Goldsmith model. This input, plotted in Figure 1, has a duration of 0.03 seconds and a peak of 190 m/s\(^2\).

The displacement response to this force was plotted using the MATLAB state space function, \texttt{ss()}, and simulated time response function, \texttt{lsim()}. The displacements for several different stiffness values were plotted and compared.

**Physical Model**

A physical model, displayed in Figure 2, was constructed to support the results discovered by the mathematical model. The model includes a simple physical representation of a cervical spine and attachable brace.

![Physical model with attached brace](image)

**Figure 2** Physical model with attached brace

The cervical spine model consisted of three vertebrae constructed from 3” PVC pipe and can be seen in detail in Figure 3. The vertebrae were shaped with overlapping curved extrusions and were connected with #10-32 machine screws. This configuration allowed for a neck-like pivot movement between each vertebra. Extension springs were placed between each vertebra to allow the head to stand on its own and stiffen the motion. A 6 x 4” PVC reducing couple was bolted to the top vertebrae
of this configuration to represent a human head. The full head and neck device was then connected to a plastic flange bolted to an aluminum alloy sheet. This aluminum alloy sheet was in turn connected to a t-slotted aluminum framing platform for stability.

Several ideas and actuator technologies were considered for the creation of the neck brace. These ideas include flexible matrix composite actuators, SMA fibers wrapped around a collar, adaptive headrests, and variable stiffness actuators. An adaptive neck brace was then chosen based on simplicity of design, and ability to be nonimpending when not needed. A scissor mechanism, displayed in Figure 4, was settled upon due to its ability to fulfill the previously stated requirements as well as the additional benefits of possessing a nonlinear force-displacement, and the ability to be tailored for specific loads by operating in different positions. In a taller and thinner position, the brace becomes stiffer and acts less stiff in a shorter wider position.

Nitinol wire was chosen as the SMA actuator for its compact size, simple actuation, and inexpensiveness. An SMA wire such as nitinol wire is typically activated by heat, which can be supplied using the wire’s natural resistance and an input voltage. Nitinol wire in particular shortens by approximately 5% when fully activated.

The brace fabrication consisted of two scissor mechanisms attached on opposing sides of the model to restrict its pivoting motion. The scissor mechanisms were made using 6” long, #10-32 connecting rods. Each rod was attached to ball joint rod ends. These joints allowed for the rods to be connected with #10-32 bolts to create a diamond shape. The left and right joints of the diamond were connected using nitinol wire as indicated in Figure 4 to stiffen the mechanism. The scissor mechanisms attached to the head and base of the model using flat aluminum extrusion brackets. The lower brackets were attached to a t-slotted aluminum framing bolted into the plastic flange at the base of the model. The upper brackets were clamped to the head of the model using a hose clamp.

To gather time response data, a PCB 303A accelerometer was attached to the head of the model to read acceleration in the z direction. The accelerometer was connected to a model 482A PCB I.C.P. power supply from which the accelerometer reading was interpreted and stored by an Agilent model 35670A Dynamic Signal Analyzer.

The physical model was first tested with no brace to determine a baseline reading. The head of the model was pulled back and released from approximately 3” and allowed to oscillate until it came to a rest. The model was then tested in the same fashion with the brace attached and an approximate 1½” deflection.

In addition to time response data, it was also desired to map a frequency response of the model. In order to extract an input force reading to correlate with the output accelerometer signal, a PCB Impact Hammer was used. This impact hammer was connected to a model 484B I.C.P. Power Unit, which was in turn connected to the same dynamic signal analyzer as the accelerometer. The tip of the hammer was covered with a rubber plug to soften the blow of the hammer and isolate the lower mode shapes of the model. To collect data, the model was lightly hit by the impulse hammer and let come to a rest repeatedly for a sample time of approximately 10 seconds. The data from the accelerometer and hammer were also stored in the dynamic signal analyzer.
Analytical Model Results

The goal of the analytical model investigation was to determine the effectiveness of the increased stiffness provided by a brace in preventing neck injury. This was done by increasing the stiffness of the neck model in the z and θ directions, corresponding to the directions most directly affected by the brace. Figure 5 displays the displacement time response of the neck, where each plot represents different brace stiffness values. The numbers above each plot represent the percentage increase of the designated stiffness terms.

![Analytical model displacement time results](image)

It is observed from the plots above that the displacement amplitude resulting from the imposed acceleration force decreases as the neck is stiffened. It is assumed that with a smaller imposed displacement, it would require less strain in the neck of the pilot to resist those displacement forces. It would be beneficial to investigate this relation in a future study.

Physical Model Results

Although nitinol wire was used to connect the two corners of the scissor mechanism, it was not activated for this study. In theory, the wire would be excited with increasing voltages as acceleration forces increased, causing the wire to shorten and stiffen the neck brace. This would allow for one to customize the amount of desired assistance provided from the neck brace in relation to different strengths of acceleration forces experienced. Upon receiving the wire, it was found that the displacement caused by exciting the wire with a voltage was not significant enough to vary the stiffness of the scissor mechanism.

The physical model time response data were used to determine the approximate natural frequencies and initial amplitudes of both the braced and non braced responses. Using these two values, a transfer function was fitted to the data by plotting step responses with varying damping coefficients and comparing those plots to the original data. The transfer function was determined by using Equation 8. Plots of each transfer function were created using the MATLAB step() function.

\[
TF = \frac{A}{s^2 + 2\xi\omega_n s + \omega_n^2}
\]

In the above equation A represents the amplitude of the function, \(\xi\) represents the damping ratio, and \(\omega_n\) represents the natural frequency. Both the estimated transfer functions and the original data are plotted in Figure 6.

![Unbraced time response and fitted transfer function](image)

![Braced time response and fitted transfer function](image)

Once the transfer functions were fitted, they were used to generate time displacement responses using the same MATLAB command. These time responses
are shown in Figure 7. As seen in the figure, the displacement of the braced system is less than the unbraced system for equivalent initial conditions.

The plots for both the non-braced and braced frequency responses are displayed in Figure 8. The non-braced model was originally sampled over a 100 Hz range, but was found to have too large a resolution to make out any resonant frequencies. The range was therefore reduced to 25 Hz.

**Figure 7** Time displacement response of the braced and unbraced models.

The frequency response is the steady state response to an input sinusoidal force. By looking at the frequency response of the unbraced model, it can be inferred that its first resonant frequency occurs at 1.5 Hz. This resonant peak is much higher than any of the readings experienced by the braced model, showing that the unbraced model will experience much higher acceleration forces for certain sinusoidal input forces. This data also supports the conclusion that a neck brace may help preserve the safety of the pilot’s neck. The data also shows that a brace adds more damping to the pilot’s neck allowing it to more quickly return to a state of equilibrium after being disturbed.

**Conclusions**

An experiment was conducted to develop a simplified analytical model of the neck, determine the effectiveness of using a brace to prevent neck injury under high acceleration loads, identify a potential brace mechanism capable of dynamically changing the stiffness of the brace, identify actuator technology capable of actuating the mechanism effectively and efficiently, and to experimentally validate the chosen mechanism and actuator. The following conclusions were drawn.

1. As determined by a mathematical model of the neck, head displacement due to acceleration forces can be minimized using an adaptive neck brace that supports the neck in both forward and backward translation, as well as up and down rotation. This minimized deflection due to acceleration would likely decrease the amount of strain experienced by a pilot’s neck when resisting this deflection.

2. A physical model was successfully used to reflect the reduction of head displacement caused by acceleration forces represented in the mathematical model.

3. The introduction of a brace avoids high resonant peaks and introduces damping to the system, allowing the pilot to return his neck to equilibrium more quickly upon being disturbed.

4. The addition of a neck brace appears to have a favorable affect on the properties that would avoid strain or injury in the necks of pilots undergoing high acceleration forces.

5. A scissor action neck brace is a possible mechanism for creating a dynamically adaptive neck brace.

6. Nitinol wire does not actuate to a great enough magnitude to be used to activate a scissor action neck brace. A smart material with much larger actuation must be used.
References


### Appendix

Table 1 Cervical spine vertebrae parameters

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<th>Ix (10^-4 kg m^2)</th>
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