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Semi-Active Control of Vertical Stroking Helicopter Crew Seat for Enhanced Crashworthiness

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I. Introduction

SHOCK load-induced injury minimization has become an important issue in helicopter seat design. Harsh vertical landings or crash landings of these aircraft tend to result in pilot or occupant spinal and pelvic injuries. The severity of injury, however, can be reduced if the vehicles are outfitted with crashworthy seat designs. Utilization of a seat suspension system to attenuate the vertical shock loads that are transmitted from the base frame of the aircraft of the vehicle and imparted into the human body is a prime factor in determining survivability [1].

Within the cockpit, energy-absorbing crew seats have greatly enhanced helicopter crash survivability. Energy absorbers (EAs) are a key component in these energy-absorbing seats. The first examples of crashworthy crew seat designs employed fixed-load energy absorbers (FLEAs) to limit an occupant's spinal load. These FLEAs are not adjustable (i.e., passive) so that their stroke and load profile are fixed at a factory-established, constant load throughout their entire operating range. Variable load energy absorbers (VLEAs) were developed subsequently to permit the occupant to manually adjust the constant stroking load by setting a dial for occupant weight. The stroking load of the VLEA is selected a priori that is proportional to the occupant weight, so that each occupant will experience similar acceleration (typically 14.5 G) and use similar stroking space during a high sink rate event. VLEAs exploit the fact that the strength of an occupant's spine is nearly proportional to occupant weight, so that the VLEA will deliver the same low injury risk regardless of occupant weight. This technology was applied in programs to retrofit new seats into platforms such as the U.S. Navy's CH-53 Sea Stallion and SH-3 Sea King aircraft [2].

Both fixed and variable load energy absorbers, however, are passive, in that they cannot automatically adapt their energy absorption or stroking profiles as a function of occupant weight, or as a function of real-time environmental measurements such as vibration level, shock level, sink rate, etc. This motivates the development of a seat suspension that uses an electronically adjustable adaptive energy absorber that can respond to such changing environmental stimuli via commands from a real-time feedback control system.

Magnetorheological energy absorbers (MREAs) offer an innovative way to achieve what is effectively a continuously adjustable profile EA [3]. Using feedback control, the MREA can smoothly adjust the load profile as the seat strokes during a crash. Thus, MREAs are expected to provide the optimum combination of short stroking distance and minimum spinal load, while automatically adjusting for the occupant weight and load level.

Of the three potential seat suspension approaches (passive, semi-active, and active), the semi-active approach is very attractive. A major drawback of a passive seat suspension based on viscoelastic or hydraulic energy absorbers is that performance is limited because neither damping nor stiffness are controllable. Furthermore, compared to active approaches, semi-active systems tend to require less power and have no stability issues (because semi-active force is always dissipative). Many researchers have been inspired to develop novel seat suspensions showing improved shock and vibration attenuation performance by permitting stiffness or damping to be adaptable and controllable. Wu and Griffin [4] examined several semi-active control algorithms for reduction in the severity of seat suspension end-stop impacts. Choi et al. [5,6] evaluated the attenuation of seat vibration using skyhook and sliding mode control algorithms on both electrorheological (ER) and MR seat suspensions for commercial vehicles. Park and Jeon [7] developed a Lyapunov-based robust control algorithm which compensates for energy absorber time delay and evaluated vibration control performance of an MR seat suspension. McManus et al. [8] investigated the use of MR seat suspensions to reduce the incidence and severity of end-stop impacts, showing impressive end-stop impact attenuation performance and reduced vibration exposure levels. Recently, Choi and Wereley [1] analyzed the biodynamic response of the human body protected by a controlled MR rotorcraft seat suspension to both sinusoidal vibration and shock loads, and showed that the MR suspension had better performance than a passive hydraulic seat suspension.

In the present study, a control algorithm is presented through which a magnetorheological energy absorber may be used in a crew seat suspension to automatically accommodate occupants of varying weight.

II. Mathematical Modeling

A nonlinear lumped parameter model of a seated occupant was coupled with the nonlinear Bingham-plastic force model for an MREA, giving the model depicted in Fig. 1. In this model, the seat, denoted by M_1, is fixed to the floor through the MREA F_{MR} and spring K_1. In addition, an end-stop buffer is implemented, which produces a nonlinear spring reaction force F_{st}, when the suspension stroke exceeds its free-suspension travel. The soft seat cushion is simply represented as a stiffness and damping (K_{2c} and C_{2c}, respectively).

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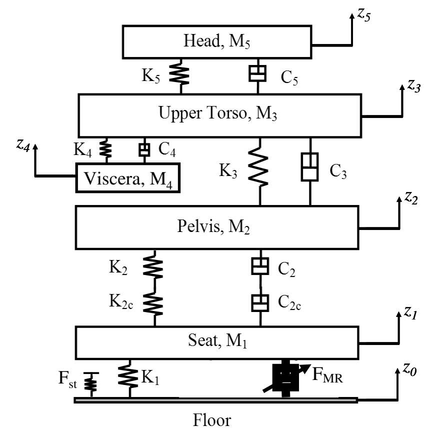

Fig. 1 Mechanical model of MR seat suspension coupled with a human body in a seated posture.

This lumped parameter model assumes the human is seated and that 29% of the body weight is supported by the feet [1,9]. The body is divided into four parts: pelvis, upper torso, viscera, and head, represented by mass M_i, stiffness K_i, and damping C_i, where $i = 2$, 3, 4, and 5, respectively. The motion of this system is governed by the following equations [1]:

$$M_1\ddot{z}_1 = -K_1(z_1 - z_0) + K_{2t}(z_2 - z_1) + C_{2t}(\dot{z}_2 - \dot{z}_1) - F_{\mathrm{MR}} \quad (1)$$

$$M_2\ddot{z}_2 = -K_{2t}(z_2 - z_1) + C_{2t}(\dot{z}_2 - \dot{z}_1) + K_3(z_3 - z_2) + C_3(\dot{z}_3 - \dot{z}_2) \quad (2)$$

$$M_3\ddot{z}_3 = -K_3(z_3 - z_2) + C_3(\dot{z}_3 - \dot{z}_2) - K_4(z_3 - z_4) - C_4(\dot{z}_3 - \dot{z}_4) + K_5(z_5 - z_3) + C_5(\dot{z}_5 - \dot{z}_3) \quad (3)$$

$$M_4\ddot{z}_4 = K_4(z_3 - z_4) + C_4(\dot{z}_3 - \dot{z}_4) \quad (4)$$

$$M_5\ddot{z}_5 = -K_5(z_5 - z_3) - C_5(\dot{z}_5 - \dot{z}_3) \quad (5)$$

where

$$K_{2t} = \frac{K_2 K_{2c}}{K_2 + K_{2c}} \quad (6)$$

and

$$C_{2t} = \frac{C_2 C_{2c}}{C_2 + C_{2c}} \quad (7)$$

In Eq. (1), z_0 is the displacement of the floor. The initial conditions for this problem are $z_i = 0$ and $\dot{z}_i = -v_0$, where $i = 0, 1, 2, 3, 4$, and 5, and v_0 is the initial vertical landing velocity (or sink rate) of the helicopter. It is assumed that the deceleration applied to the floor due to the sink rate is in the form of a triangular pulse with 51 ms duration.

The stiffness of the pelvis K_2 is modeled by the nonlinear function [10]:

$$K_2 = 8.1075e7(z_2 - z_1)^2 \quad (8)$$

The stiffness of the upper torso is also nonlinear [10]:

$$K_3 = \begin{cases} 3.78e6 + 1.09e7(z_2 - z_3) - 2.69e7(z_2 - z_3)^2, & \text{if } (z_2 - z_3) \geq 0.04 \\ 77{,}044, & \text{if } (z_2 - z_3) < 0.04 \end{cases} \quad (9)$$

Table 1 Preliminary MR seat suspension parameters

Quantity	Symbol	Value	Units
Mass of seat	M_1	11.5	kg
Mass of pelvis	M_2	29	kg
Mass of upper torso	M_3	21.8	kg
Mass of viscera	M_4	6.8	kg
Mass of head	M_5	5.5	kg
Stiffness of coil spring	K_1	0.0	kN/m
Stiffness of soft seat cushion	K_{2c}	37.7	kN/m
Stiffness of viscera	K_4	2.84	kN/m
Stiffness of head	K_5	202.3	kN/m
Cushion damping	C_{2c}	159	N · s/m
Pelvis damping	ζ_2	0.25	——
Torso damping	ζ_3	0.11	——
Viscera damping	ζ_4	0.5	——
Head damping	ζ_5	0.1	——

The damping C_i is given by

$$C_i = 2\zeta_i\sqrt{M_i K_i} \qquad \text{if } i = 2, 3, 4, 5 \quad (10)$$

where ζ_i is the damping ratio of each part of the human body. Because K_2 and K_3 are nonlinear functions, C_2 and C_3 are also nonlinear. Lastly, the nonlinear spring reaction force F_{st} due to the end-stop buffer is given by [1]

$$F_{\mathrm{st}} = 8.0e^4[z_1 - z_0 - z_{\mathrm{st}}\mathrm{sgn}(z_1 - z_0)] + 3.4e8[z_1 - z_0 - z_{\mathrm{st}}\mathrm{sgn}(z_1 - z_0)]^3 \quad (11)$$

The parameters of the MR seat suspension model used for this preliminary study are specified in Table 1 [1,9,10]. The biodynamic parameters represent those obtained from dynamic tensile tests by Liu et al. [9,10] for an approximate 90th percentile male. When considering varying occupant weights, these mass values are scaled linearly by total occupant weight. In addition to these masses, 25 lbs of body worn equipment has been added to the upper torso, and 5 lbs have been added to the head to account for a helmet. Based upon studies performed by Boileau et al. [11], the biodynamic stiffnesses have been assumed to remain constant over varying occupant weights. Lastly, it should be noted that the coil spring stiffness K_1 has been set to zero. This is because for a crash event, an energy storing device (spring) will produce an undesirable rebound lumbar load.

In the early 1970s, analysis and testing were performed to determine a limit load for seat energy absorption systems, that is, the load at which an FLEA would start and continue to stroke. Relating the seat performance to tolerance data and cadaveric testing, an energy-absorbing limit load of 14.5 G was determined to keep the load duration in the humanly tolerable range. This means that the FLEA would stroke at 14.5 times the effective occupant weight (% of body weight not supported by floor plus weight of body worn items) plus the weight of the stroking portion of the seat [2,12].

Given this, an MR VLEA seat system (that is, an MR seat energy absorption system which varies its limit load based upon occupant weight) is easily realized. A simple controller can be designed to determine the effective occupied seat mass M_{eff} by a weight sensor in the seat, static deflection of the seat, or via manual setting. Once this value is determined, the controller then adjusts the limit load F_L as

$$F_L = \alpha_{\mathrm{limit}} \times M_{\mathrm{eff}} = \alpha_{\mathrm{limit}} \times K_1(z_1 - z_0)_{\mathrm{static}} \quad (12)$$

The limit factor α_{limit} is the injury tolerance criteria of choice (in G). In this study, the 14.5 G criteria discussed above is used. The control can then modulate MREA yield force real time to keep the damper force at this constant limit load. MREA yield force modulation may be performed using a simple load tracking control algorithm or by using the MREA force model. In this study, the Bingham-plastic force model is used as an example; therefore the yield force is varied using the following simple equation:

$$F_y = F_L - C(\dot{z}_1 - \dot{z}_0) \quad (13)$$

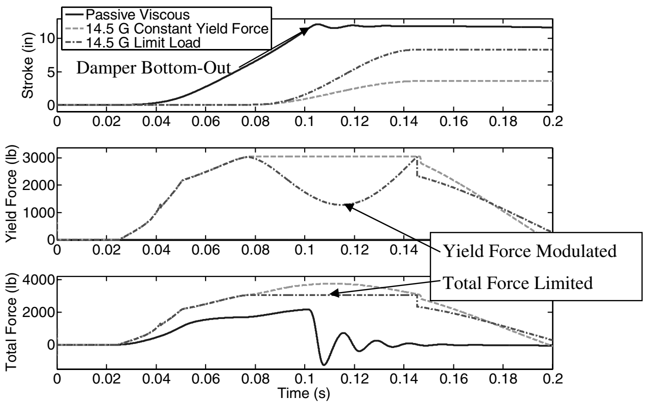

Fig. 2 MR damper time response for passive viscous, constant yield force, and load-limiting control.

where F_y is the MREA yield force, F_L is the load limit, and C is the MREA viscous damping. Knowing the instantaneous velocities of the seat and floor, the controller then uses Eq. (13) to determine the desired MREA yield force. This MR yield force modulation equation can become more complex if other MREA or MR damper force models are used (those including hysteresis, compressibility, etc.).

III. Results and Discussion

Figure 2 shows the resulting time response of the MREA from a 42 ft/s sink rate crash assuming a symmetric triangular acceleration pulse with 51 ms duration, 11.5 in. of available stroke, and a 95th percentile male occupant (212 lbs) [12]. The top plot shows that, in the passive, viscous-only case, the MREA bottoms out quickly, which leads to an increased reaction into the occupant's spine. In the constant yield force case, the total damper force (bottom plot) quickly increases to the 14.5 G yield force setting. As piston velocity increases, the total MREA force further increases beyond the 14.5 G load due to the viscous force component. The load-limiting control, however, prevents the total force imparted into the seat (bottom plot) from exceeding the limit value by modulating the yield force (middle plot). It can also be seen that limiting the load to 14.5 G efficiently uses more of the available stroke than the constant yield force case.

Figure 3 shows these MREA time response plots for load-limiting control and varying occupant weight. It can be seen that the load-limiting control effectively limits the load imparted into the seat to the respective 14.5 G level for the 5th percentile female (102.8 lbs) as well as the 50th and 95th percentile males (171 and 212 lbs), respectively [12]. It can also be seen that the controller prevents the MREA from bottoming out in each of these cases. Additionally, note that the peak lumbar loads predicted by the biodynamic model are 1702, 1440 , and 930 lbs for the 95th percentile male, 50th percentile male, and 5th percentile female, respectively. These are well below the respective limits of 2534, 1610, and 1281 lbs published in [13].

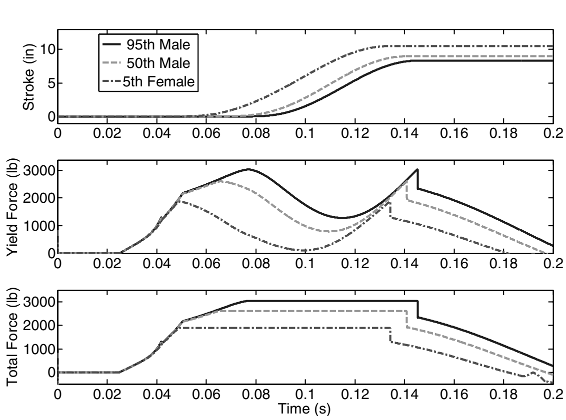

Fig. 3 MR damper time response for load-limiting control and varying occupant weight.

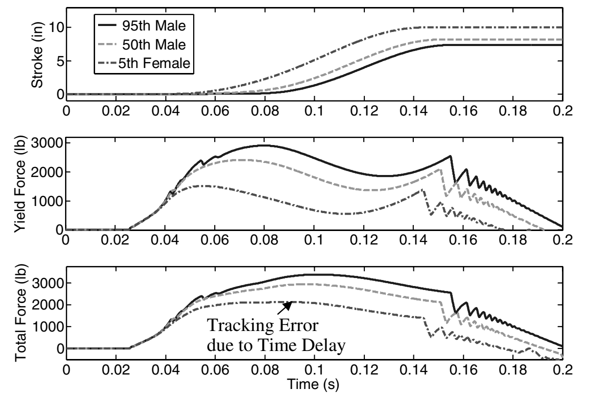

Fig. 4 Time response of MR damper for load-limiting control and 10 ms time constant.

The previous simulations assume an ideal MREA and control system, so that the practical effect of delay in the MREA time response is now considered. To emulate this time response, the control action, or desired damper force f_d, is assumed to pass through a first-order low-pass filter given by

$$\dot{f}_{d_f} = \frac{f_d - f_{d_f}}{\tau} \tag{14}$$

where f_{d_f} is the filtered control input and τ is the time constant [1,14]. Choi and Wereley [14] have experimentally calculated the response time of their MR dampers (or MREAs) to be 7–8 ms. Assuming similar MREA performance, the simulation of Fig. 3 was rerun with a 10 ms time constant, giving the results shown in Fig. 4. First, note that some oscillations have appeared at the onset of the crash event (0.025 s) as well as once the MREA has finished stroking (0.140 s). These are due to the dynamics of the coupled nonlinear biodynamic model and are more pronounced because of the energy absorber time delay. Because of this time delay, the total force (third plot) is no longer perfectly limited to the 14.5 G stroking load. Alternatively, a slight tracking error has appeared that increases this force beyond its desired limit established in Fig. 3. The peak force values increase beyond the load limits by 10.8, 12.2, and 12.4% for the 95th percentile male, 50th percentile male, and 5th percentile female, respectively. The root-mean square (rms) of the tracking errors are 7.3, 8.1, and 10.0%, respectively. These tracking errors are likely acceptable for practical application as they are likely well within the realm of error for the 14.5 G injury tolerance criteria. Furthermore, predicted lumbar loads of 1735, 1466, and 943 lbs are still well below the limits for the 95th percentile male, 50th percentile male, and 5th percentile female, respectively.

To further test the sensitivity of the MREA time response, simulations using a 20 ms time constant were also performed (Fig. 5). It can be seen in Fig. 5 that the tracking error has not changed significantly. For this case, the peak force values increase beyond the

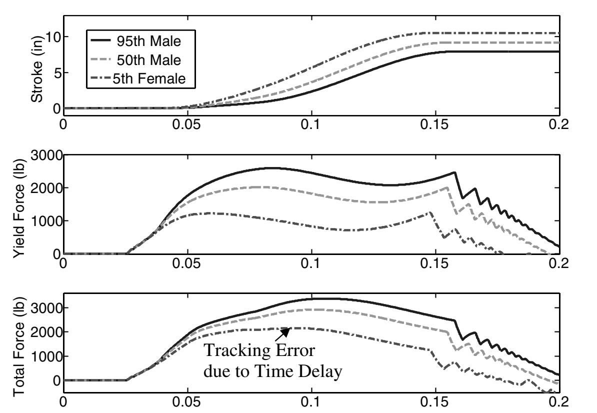

Fig. 5 Time response of MR damper for load-limiting control and 20 ms time constant.

load limits by 10.6, 11.3, and 14.0% for the 95th percentile male, 50th percentile male, and 5th percentile female, respectively. The rms tracking errors are 7.1, 7.6, and 10.4%, respectively. These tracking errors have not changed significantly from this increase in the MREA time delay. This shows that MREA time constants less than 20 ms, which are realizable in practical MREAs or MR dampers [14], should not hinder practical implementation.

IV. Conclusions and Future Work

In this study, a simple load-limiting control algorithm for an MR suspension applied to a helicopter crew seat was presented and simulation results were shown. This control algorithm realized MREAs as VLEAs, where the limit load varied automatically based on a measured (or manually set) occupant weight. Additionally, it was shown that a lumped parameter biodynamic model coupled with the MR seat suspension can be used to predict human body loads and that the lumbar loads were below tolerance limits when using this control algorithm. Last, the effects of MREA response time as long as 20 ms were shown. Although the response time slightly degraded controller performance, the tracking error was small and predicted lumbar loads remained below tolerance limits.

Future work will include the further development of semi-active controllers for crashworthiness to account for varying crash sink rate and optimize stroke vs body loads, a methodology for crashworthy MR seat suspension design, vibration control development for normal operating scenarios, and experimental verification.

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