

DRIVE BELT TENSION AS A DETERMINANT OF ROLLER PUMP HEAD SLIPPAGE

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An experience with roller pump head slippage which resulted in a reduction of blood flow during a clinical cardiopulmonary bypass has led us to investigate the potential causes of this problem. The discovery of insufficient drive belt tension in a pump module which had recently undergone repair and testing warrants a change in the routine maintenance procedures for roller pump equipment. The rationale for recommending a determination of drive belt torque tension as a preventive maintenance procedure will be discussed.

CASE REPORT

The patient was an adult 86.3 kg male, undergoing triple aorto-coronary bypass surgery. The superior and inferior vena cavae were cannulated with 34 French and 36 French (U.S.C.I.,* *A Division of C.R. Bard, Inc.*) drainage cannulae, respectively. Arterialized blood was returned through a 24 French aortic perfusion cannula** (*Argyle-Sherwood Medical Industries*) by a Sarns 6002*** occlusive modular pump with $\frac{5}{8}$ " \times $\frac{1}{8}$ " silicon pump head tubing. Temperature was controlled and oxygenation provided by a Bentley BOS 10 oxygenator.

The pump module used for arterial perfusion had recently been returned from routine overhaul and maintenance by the manufacturer. It had been inspected for electrical defects, and it had been subjected to a bench run at $\frac{1}{2}$ relative speed for four hours at zero load before clinical use. During the course of cardiopulmonary bypass the pump head was noted to be slipping, resulting in decreased blood flow in the arterial line. The flow deficit was corrected by increasing the speed of the pump head, thereby maintaining flow, despite continued slippage. The case was successfully completed with adequate extracorporeal perfusion.

COMMENT

Subsequent bench testing revealed that when the pump head developed a torque of 15 inch-pounds or less, the belt drive reduction system began to slip. Discussions with

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the manufacturer revealed that proper adjustment of the drive belt tension would allow the pump head to develop a torque of 55 inch-pounds. Testing verified that this torque setting is sufficient to provide accurate perfusion rates, since output would not depend upon the resistance in the circuit if the pump head is occlusive at 55 inch-pounds torque. A survey of Sarns 6002 modular pumps indicated that 50% had a reduced drive belt tension. These pump heads produced torques in the range of 30–50 inch-pounds.

There are many potential causes of incorrect drive belt tension such as, relaxation of belt drive with normal usage, improper pump head occlusion and damage during transportation. Pump head slippage can occur under these circumstances despite adherence to the recommended routine for servicing. While it is obvious that grossly maladjusted belt tension would cause slippage, the critical level has not been determined. The following experiment was conducted in an effort to determine the effect of drive belt tension on pump head function.

MATERIALS AND METHODS

A perfusion circuit was assembled which was identical to that used for clinical perfusion. The "patient" load was simulated with 60 cm lengths of Tygon® 5-50* Class VI tubing. Tubing with internal diameters and wall thicknesses $\frac{1}{2}'' \times \frac{3}{32}''$, $\frac{3}{8}'' \times \frac{3}{32}''$, and $\frac{1}{4}'' \times \frac{1}{16}''$ respectively were used to allow total resistance to flow to be varied. The flow through the circuit was measured with a Satham Flow Probe Sensor (SP 7519-375-604) and Satham**** Blood flowmeter (SP 22201) distal to the resistance tubing. In some experiments the pressure drop across the resistance was measured with two Satham 267 pressure transducers. The pump head occlusion was adjusted to be "just occlusive" (1) and was maintained throughout the study. Pump head speed was held constant.

Using the system described, flow was measured for each of the simulated patient loads as the drive belt tension was reduced in increments from 55 inch-pounds. Tension adjustments were made by turning the double nut located at the bottom center of the motor base (Fig. 1). To tighten, the nuts are turned counter-clockwise. To loosen they are turned clockwise(2). The belt drive tensions were calibrated with a Sturtevant/Richmont Torque Wrench (Model F-100-1) (Fig. 2). To establish torque calibrations, the wrench was placed over the central shaft (Fig. 3) of the modular pump. While holding the wrench firmly in position the pump speed was slowly increased until slippage occurred. At this point, the torque reading was recorded. Accordingly, we commenced at 55 inch-pounds of torque and proceeded to decrease the torque to 45, 40, 30, and 15 inch-pounds, respectively. These torque settings were established during perfusion for all tubing sizes.

RESULTS

The results are illustrated in Fig. 4. If the resistance in the largest tubing ($\frac{1}{2}''$), is considered to be 1.0, then, since resistance is inversely proportional to the fourth power of the radius, $(1/r^4)^{(3)}$, the resistances in the other tubing sizes are 3.16 ($\frac{3}{8}''$) and 16 ($\frac{1}{4}''$). The flow rate was 4100 ml/min at all resistances when the drive belt tension was set at 55 inch-pounds. Reduction in drive belt tension below 40 inch-pounds diminished flow

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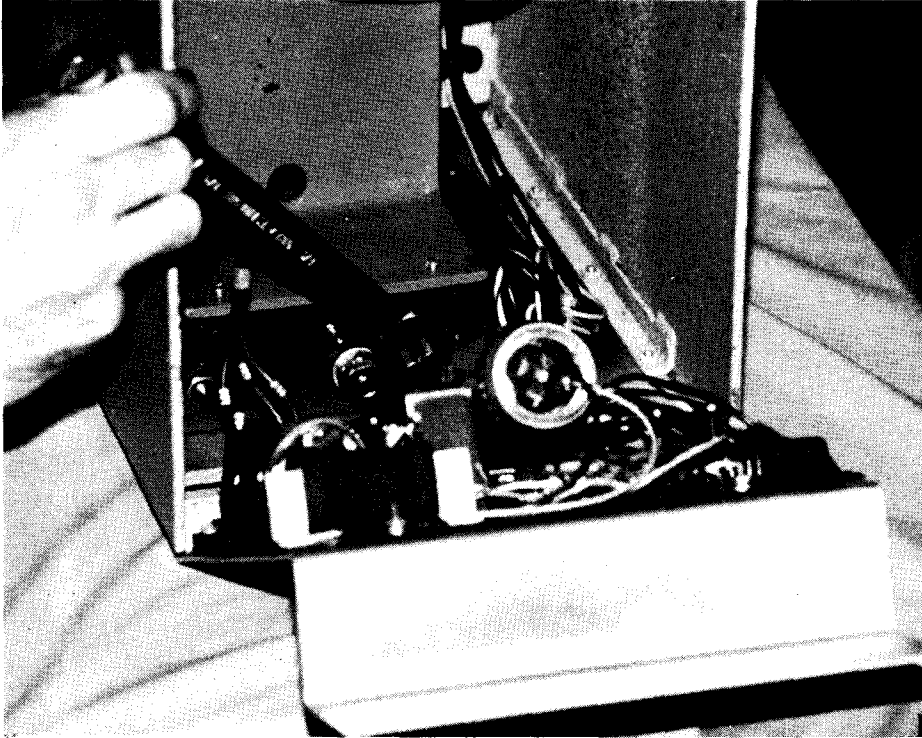


Figure 1. Drive belt tension is increased by counter-clockwise turn of double nut at motor base.

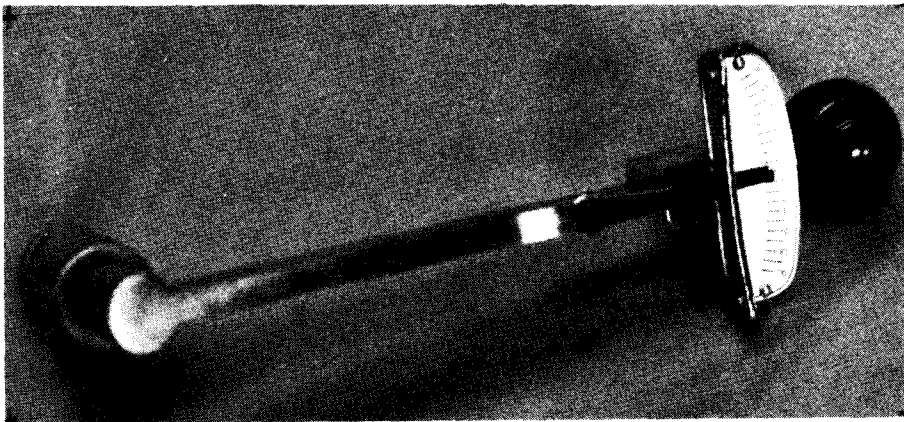


Figure 2. A Sturtevant/Richmont Torque Wrench Model F-100-1 was used to calibrate pump head torque as developed by drive belt tension.

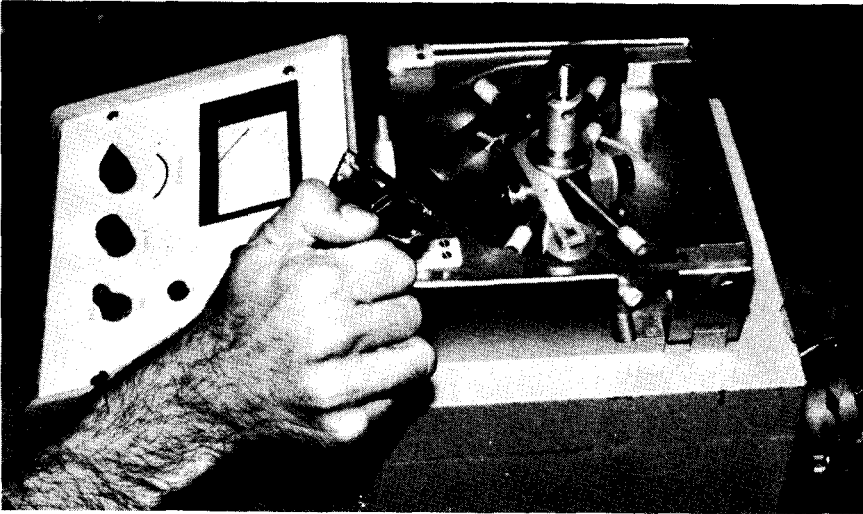


Figure 3. With torque wrench applied to the central drive shaft, maximum tension developed before drive belt slippage is recorded.

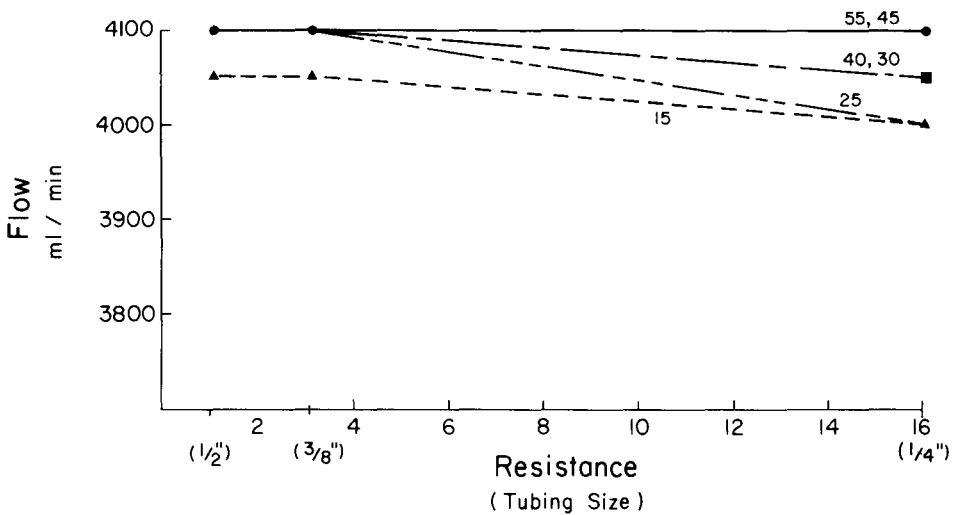


Figure 4. At roller head speed calibrated to deliver 4100 cc/min. reduction in flow was observed when torque was reduced below 40 in-lbs against higher resistance levels. At 15 in-lbs flow was reduced at all levels of resistance, and flow dropped precipitously when torque tension was further reduced.

through the highest tubing resistance while further reduction to 15 inch-pounds caused a decrease in flow at all resistance levels.

DISCUSSION

As expected, a gross maladjustment of drive belt tension, producing a torque tension of 15 inch-pounds, caused a decrease in flow at all resistances. The finding that more moderate maladjustments did not lead to flow reduction except at the higher resistances gives rise to two questions. First, are the resistances used in this study comparable to those encountered during a clinical perfusion? Second, are the observed reductions in flow of clinical significance? In an attempt to answer these questions, the following thesis is proposed. Actual resistance in the experimental perfusion system is defined as pressure drop over the length of the tubing divided by flow. When this calculation was made with a 60 cm length of $\frac{1}{4}$ " tubing at a flow rate of 6.6L/min., a pressure drop of 280 mm/Hg. was observed and a resistance of 2.5 R units was calculated. The total peripheral vascular resistance of a healthy human is 1.0 to 2.0 R units⁽⁴⁾, and may be greater than 4.0 R units in patients with vascular disease or low cardiac output⁽⁵⁾. The resistances used in this experiment approximate normal values, and are very likely lower than those seen in pathological states.

The decrease in flow observed in this study is unlikely to be clinically significant. However, the study was designed to test the performance of the pump under light to moderate loads. A number of factors might be encountered during a clinical perfusion which would increase the "patient" load. These studies were conducted with the pump head "just occluded." Over occlusion would increase the load on the belt drive. This study was conducted with water rather than blood; the higher viscosity of blood will increase the load. Finally, this study was conducted with a smooth length of tubing with no branchings or constrictions which might cause turbulence. Clinically, turbulence is induced when such branchings, constrictions, and convolutions occur, especially with small cannulas and tubing. This can significantly increase the resistance of the system and the load on the belt drive.

CONCLUSIONS

We have found that through use or improper adjustment, it is possible for the belt drive reduction system of the Sarns 6002 occlusive modular pump to fail to develop the recommended torque of 55 inch-pounds. Gross maladjustment will produce visible slippage and decreased perfusion. With more subtle maladjustments, which might result from normal wear, while slippage may not be obvious, a small decrease in perfusion can occur at normal loads, whereas, high loads may cause a considerable decrease in flow under these circumstances. Calculated flow rates depend upon relative torque settings. The results of this study suggest that a torque setting of 35 to 45 inch-pounds will deliver predictable flow rates under most clinical conditions while allowing slippage when high line resistance occurs: e.g., kinking or accidental clamping. The setting of 55 inch-pounds might result in line rupture under high loads. The delivery of efficient and accurate perfusion by roller pump equipment requires frequent determination and adjustment of drive belt tension in the Sarns 6002 roller pump module.

REFERENCES

1. Meyer, M. M.: Blood Pump Occlusion. *J. Extra-Corporeal Technology* V (4). 8-15, 1973.
2. Instructional Manual: "Heart Lung Modular Equipment" Section 2.11, Sarns Inc., Ann Arbor Michigan 48163. Copyright 1969.
3. Spencer, M. P. and Denison, A. B.: Pulsatile Flow in the Vascular System. *Handbook of Physiology II* (2), p. 841, American Physiological Society, Washington, D.C. 1963.
4. Ganong, W. F.: *Review of Medical Physiology*, 7th Ed. Lange 1975, p. 424.
5. Guyton, A. C.: *Textbook of Medical Physiology*, 5th Ed. W. B. Saunders Co., 1976, p. 229-230.