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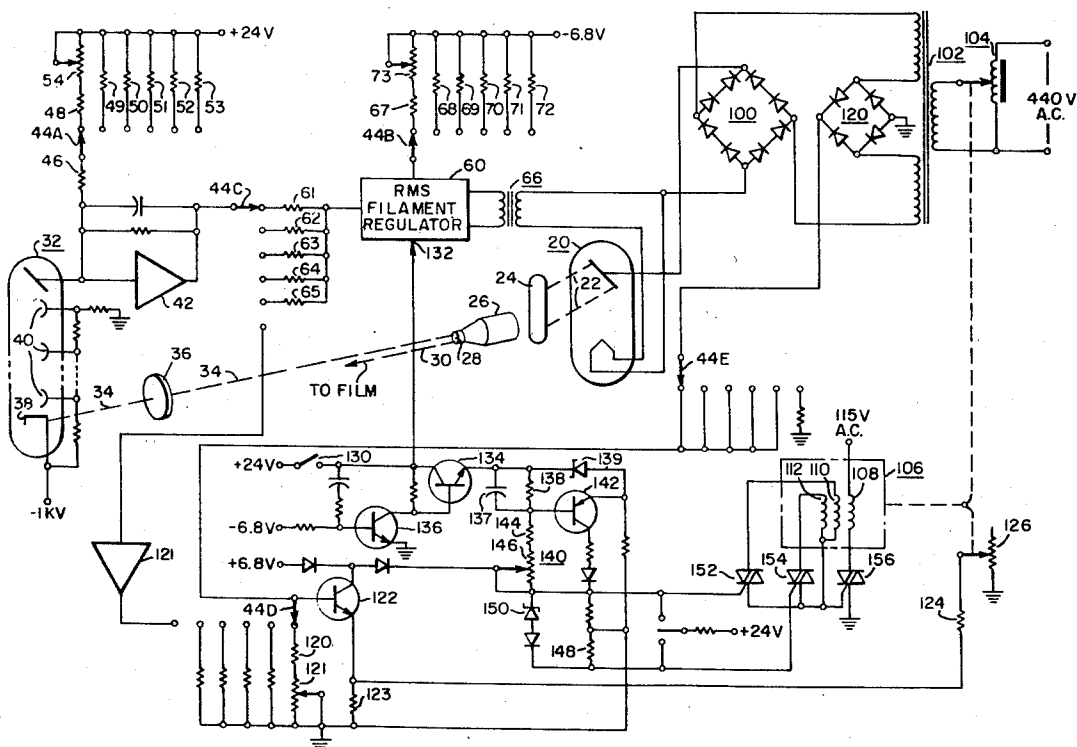
[50] Field of Search..... 250/65, 95,
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[56] **References Cited**
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[54] **BRIGHTNESS STABILIZER WITH IMPROVED
 IMAGE QUALITY**
13 Claims, 2 Drawing Figs.

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250/103
 [51] Int. Cl..... **H05g 1/36**

ABSTRACT: A brightness stabilizer which permits the radiologists to operate X-ray apparatus in fluoroscopic modes or in other modes of operation where an image intensifier is used, such as cine or 70-mm. spotfilms, in such a way as to expose the patient to the lowest level of radiation possible as a function of the type of X-ray image required.



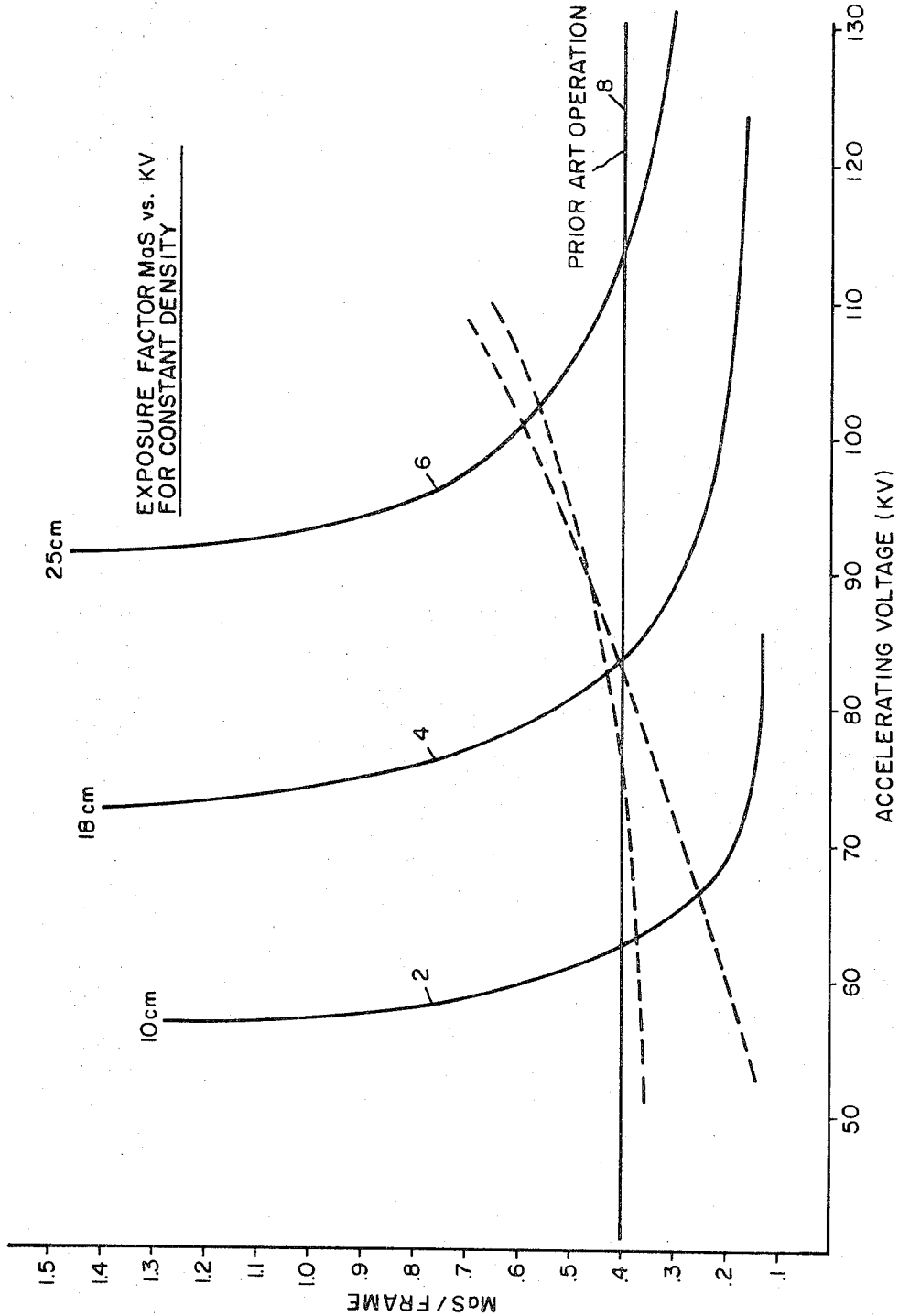


FIG. 1.

WITNESSES:

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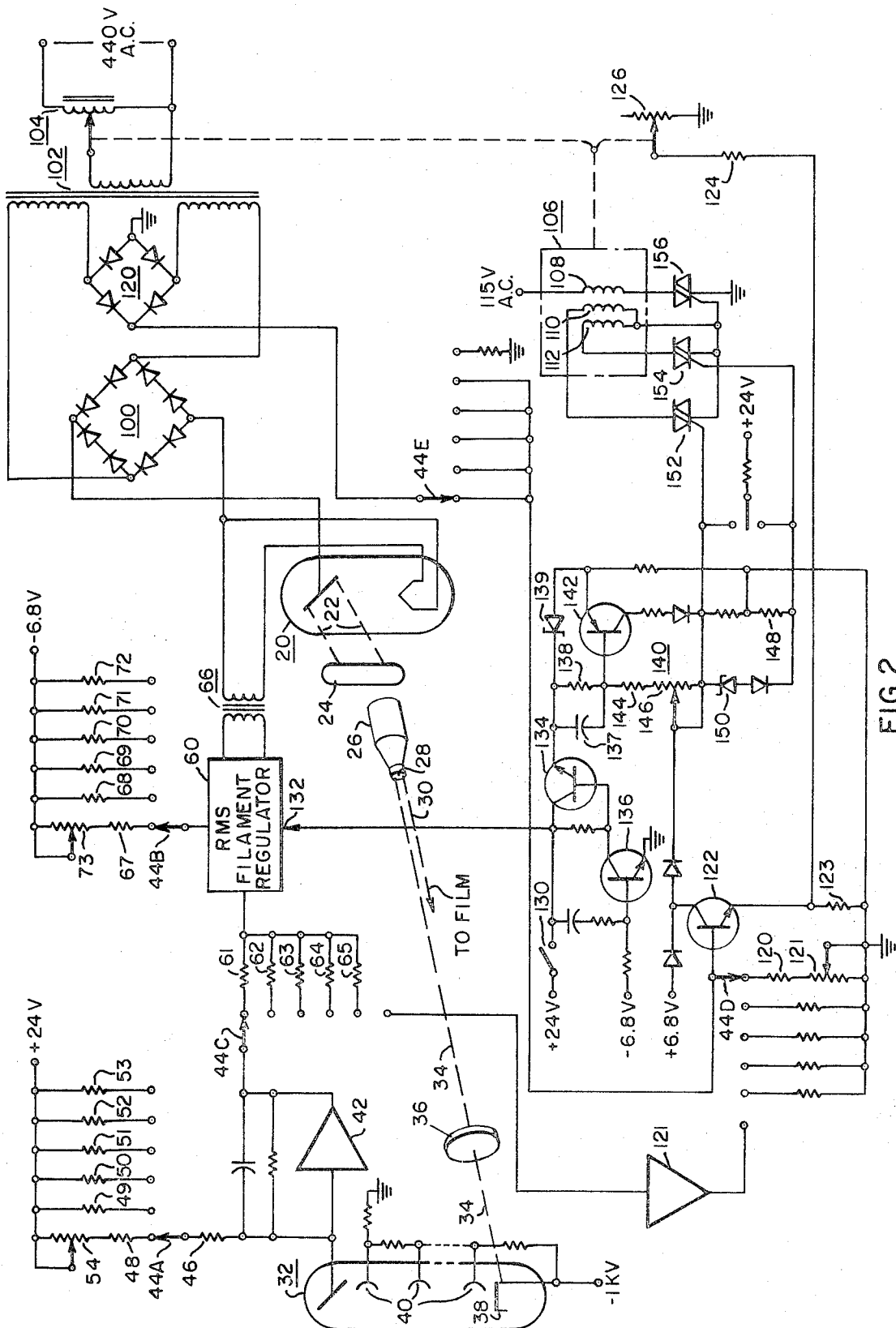


FIG. 2.

BRIGHTNESS STABILIZER WITH IMPROVED IMAGE QUALITY

CROSS REFERENCES TO RELATED APPLICATIONS

The present invention may be utilized with other circuitry for controlling an X-ray generator, for example, as described and claimed in copending patent application Ser. No. 742,463, filed July 3, 1968, "An RMS Current Regulator" by Melvin P. Siedband and Jack L. James; copending patent application Ser. No. 860,603, filed Sept. 24, 1969 entitled "X-Ray Tube Control Circuitry" by Melvin P. Siedband, Philip A. Duffy and Jack L. James; copending patent application Ser. No. 860,686, filed Sept. 24, 1969 entitled "Starting Voltage Suppressor Circuitry for an X-ray Generator" by Fred J. Euler and Jack L. James; and copending patent application Ser. No. 28,665, filed Apr. 15, 1970 entitled "Heat Sensing Circuit" by Melvin P. Siedband and Jack L. James; all being assigned to the present assignee.

BACKGROUND OF THE INVENTION

1. Field of the Invention

B. The present invention relates generally to X-ray generators and more particularly relates to a brightness stabilizer with improved image quality.

2. Description of the Prior Art

X-ray images may be evaluated as a function of the characteristics of the noise distribution of such images. Since X-ray images are formed as a result of the scintillation produced by individual X-ray quanta, the minimum average number of X-ray quanta at the image detector can be calculated. Sufficient brightness to cause film darkening is usually present for X-ray generators with modern image intensifier tubes. However, the object of the good image intensifier-photographic film system is not simply to darken film but to obtain the proper photographic information. The problem therefore is to set the X-ray factors to obtain the appropriate number of detectable X-ray quanta at the input to the image intensifier for a picture of acceptable noise level sufficient to produce the appropriate resolution, gray scale and dynamic range for that image format.

Thus far, the best pictures have been taken by determining a reasonable value of X-ray beam current and stabilizing the operation of the X-ray generator at that value by adjustment of the X-ray tube accelerating voltage for constant brightness. The X-ray generator may have many modes of operation and the beam current is chosen as a function of the cine frame rate. For example, an operating mode of 15 frames per second may require average beam currents of 10 milliamperes for proper pictures. If the patient is of small build, accelerating voltage of 60 kv. may be used, whereas a rather robust patient may require accelerating voltage in excess of 110 kv. Prior art systems usually work in the following manner: a starting point voltage is chosen by programming the control as a function of the patient's size; that is, small, medium, or large. The beam current of the X-ray tube is selected as a function of the operating mode such as 10 milliamperes at 15 frames per second of the camera, 20 milliamperes at 30 frames per second, etc. The cine camera is optically coupled to the output phosphor of the image intensifier while a photomultiplier tube also senses the average light level of the output phosphor. If the brightness of the output image is not sufficient as directed by the photomultiplier, the beam current of the X-ray tube is automatically adjusted to either a higher or lower value. Again, considering the operating mode of 15 frames per second, the bounds on beam current may be set at 8 and 12 milliamperes such that if the beam current exceeds these bounds a motor-driven system responsive to the brightness of the image intensifier causes the accelerating voltage to change by adjustment of the motor-driven autotransformer.

It has been observed however, that operation at higher accelerating voltage gives rise to excess noise, probably as a result of the modulation of the scattered X-ray photons. What-

ever the case, it has been shown that if the brightness of the image is maintained constant by covariation of the X-ray attenuation, that is, changes in the thickness of the patient and varying accelerating voltage, images at the higher accelerating voltages are indeed noisier than images at the lower voltages.

Another problem arises as a result of the design characteristics of the high-voltage transformer which feeds the X-ray tube. It is well known in the art that a small change of X-ray accelerating voltage results in a rather substantial change in the penetration of the beam and hence a rather substantial change in the brightness of the image. X-ray transformers inherently have a high level of voltage regulation. Under extreme conditions, the output voltage of the transformer may fall by 35 percent of its open circuit voltage. As a result, an increase in X-ray beam current may cause increased regulation of the high voltage transformer thereby decreasing the total image brightness. The brightness stabilizing servo would undergo a change in the sensing direction. Depending upon the design of the particular transformer, sense reversal may occur at current levels as low as 60 milliamperes or as high as 150 milliamperes. For the radiologist photographing heavy patients at high rates, say 60 to 120 frames per second, it is possible for the system to require high values of X-ray beam current. If the sense of the system is reversed when operating in these modes, the system cannot stabilize itself.

Therefore an object of the present invention is to provide an improved image-stabilizing system wherein for certain operating modes the system is preset to avoid sense reversal of the stabilizing circuitry.

It is an object of the present invention to provide an improved brightness stabilizer wherein the image will be of optimal quality.

Another object of the present invention is to provide an improved image quality which will selectively emphasize control of X-ray quanta by means of the anode current rather than the accelerating voltage.

Another object of the present invention is to provide a brightness stabilizer which is noise limited as the accelerating voltage is increased for the lowest possible X-ray factors.

Another object of the present invention is to provide a brightness stabilizer capable of performing under various operating modes.

It is a further object of the present invention to provide a brightness stabilizer capable of operation in various fluoroscopic and cine radiographic modes.

SUMMARY OF THE INVENTION

Briefly, the present invention accomplishes the above-cited objects and other objects and advantages by providing an improved brightness stabilizer wherein the accelerating voltage is controlled in response to the magnitude of beam current when outside of a predetermined range. Further, when the accelerating voltage is increased to a region where excess noise will deteriorate the image, the predetermined range of anode current is translated upward thereby allowing the accelerating voltage to be lowered. Should the operating mode be such that the required X-ray beam current causes a decrease in total image brightness as a result of the increased regulation of the high-voltage transformer the brightness stabilizer can be reset to a fixed range of beam current and the accelerating voltage varied as a function of the brightness of the image.

BRIEF DESCRIPTION OF THE DRAWING

Further objects and advantages of the present invention will be readily apparent from the following detailed description taken in conjunction with the drawing in which:

FIG. 1 is a graphical illustration of exposure factor curves for constant density which are useful in understanding the operation of the present invention; and

FIG. 2 is an electrical schematic diagram of an illustrative embodiment of the present invention.

DESCRIPTION OF THE PREFERRED EMBODIMENT

Referring to FIG. 1, constant brightness curves 2, 4, and 6 have been shown as a function of patient size; for example, 10 cm., 18 cm., and 25 cm. One may adjust the anode current and hence milliamp second exposure per frame or the accelerating voltage to operate along a constant brightness curve which will lead to proper film exposure. From FIG. 1 it can be seen that as the size of the patient varies so must the accelerating voltage to provide proper film exposure. Further, the mode of operation of X-ray generator may determine necessary beam current as a function of the frame rate of the camera; the greater the frame rate the greater the ampere current.

Prior art operation is illustrated by the curve 8 where, it can be seen, the accelerating voltage is varied with the size of the patient. For a predetermined mode of operation at a given anode current the accelerating voltage was varied in accordance with the size of the patient. At higher accelerating voltages however, the resulting images are noisier than those obtained at the lower voltages.

When operating in accordance with the present invention the stabilizing point or range of the beam current is adjusted as a function of the accelerating voltage to maintain constant noise level. In other words, when taking pictures at 15 frames per second operating mode for a thin patient an accelerating voltage of 60 kv. may permit stabilization at 10 milliamperes, whereas taking pictures of a more robust patient at 15 frames per second and 100 kv. may dictate the use of 20 milliamperes beam current.

Should however the anode current be increased to a point where sense reversal occurs as a result of increased regulation of the high-voltage transformer, then the operating mode of the present invention will simply set the anode current to a predetermined value and vary the accelerating voltage only in response to the brightness of the image.

Referring to FIG. 2, an X-ray tube 20 emits X-ray quanta 22 which passes through a target or patient 24 forming an image on the intensifier tube 26. The output phosphor 28 provides illumination 30 to the film while at the same time a photomultiplier 32 views the brightness of the output phosphor 28. The light 34 passing through a lens 36 strikes the cathode 38. Dynodes 40 enhance the current within the photomultiplier 32. The photomultiplier 32 feeds an output current proportional to the intensifier brightness to the input of an operational amplifier integrator 42. A mode selector switch 44A determines a bias or reference current through resistor 46 to which the photomultiplier current is compared. The mode selector switch 44A through E is a gang-operated switch which selects the operating mode of the X-ray generator and determines the sensitivities, operating points, and brightness requirements of the stabilization system as required for the different operating modes such as fluoroscopy and cine at film speed, say 7.5 frames per second, 15 frames per second, 30 frames per second, 60 frames per second, and 120 frames per second. The higher the brightness of the intensifier 26 the greater the bucking current for compensation that will be required. As higher brightness is required for a greater film rate, the choice of resistors 48 through 53 is made to increase the bucking current. The system is shown with the switch 44A in the fluoroscopic position so that only a small reference current is provided by the setting of variable resistor 54.

At optimum brightness for the fluoroscopic mode of operation, the output of the operational integrator 42 is zero and no correcting current will be fed to the RMS filament regulator 60 through the switch 44C. The relative gain between the operational amplifier 42 and the RMS filament regulator 60 is determined by the value of the series resistor 61 through 65 selected by the operating mode switch 44C. In the fluoroscopic mode of operation, the gain is determined by the magnitude of input resistor 61.

The filament regulator 60 may be of any suitable type, an RMS regulator type being illustrated. Such a filament current

regulator is more fully described and claimed in copending patent application Ser. No. 742,463, filed July 3, 1968, entitled "RMS CURRENT REGULATOR" by Siedband and James and assigned to the same assignee. The regulator 60 is such that during X-ray off conditions the output of the operational amplifier integrator 42 is disconnected from the comparison circuits within the regulator 60. When operating, the filament regulator 60 will control filament current through the filament transformer 66 to the X-ray tube 20.

More particularly, the starting point for filament current of the X-ray tube is selected by the mode selector switch 44B. The filament current at the starting point for the various operating modes are determined by the values of resistors 67 through 72. A starting point for fluoroscopic current therefore will be determined by the setting of the variable resistor 73 when the mode switch 44B is in the position indicated.

If the output current of the photomultiplier 32 indicates inadequate light, the output voltage of the operational amplifier integrator 42 moves in a negative direction providing a command signal to the filament regulator 60 to increase the output to the filament transformer 66 thereby causing the filament current of the X-ray tube 20 to increase. Increased filament current will then tend to raise the beam current and therefore the brightness of the image.

The accelerating voltage for the X-ray tube 20 is provided by a main high-voltage rectifier 100 which is fed by a high-voltage transformer 102, which in turn is fed by means of an autotransformer 104. The accelerating voltage for the X-ray tube 20 is therefore determined by the position of the autotransformer 104. The autotransformer 104 is driven by a motor 106 including a main winding 108, a clockwise winding 110, and a counterclockwise winding 112.

A current metering bridge 120 monitors the actual beam current of the X-ray tube 20. When the mode selector switch 44E and 44D is in the fluoroscopic position, the beam current of the X-ray tube 20 is viewed across resistors 120 and 121 and seen by the base of transistor 122 connected in a current amplifier configuration.

Resistors 120 and 121 are chosen such that the average voltage drop thereacross will be say 5 volts at a current level related to the values of the two series resistors. If the resistors are initially set for a value of 5.0 volts peak at 2 milliamperes and assuming for the moment that resistor 124 and motor-operated variable resistor 126 is not in the emitter circuit of the transistor 122, the collector current of the transistor 122 will be directly proportional to the X-ray tube beam current. With different modes of operation as determined by the position of the switch 44D, the gain of the current amplifier 122 will be preselected in accordance with the predetermined range over which the anode current will be allowed to vary. Should the anode current vary outside of the predetermined range the circuitry including and following the current amplifier 122 will energize the drive motor 106 to control the anode voltage to the X-ray tube 20.

During startup, a switch 130 connects a control voltage illustrated as +24 volts, to enable the filament regulator 60 with an input at 132. After a time delay appropriate to allow the beam currents to stabilize the 24 volts is connected across a voltage divider circuit 140. Transistors 134 and 136 plus their associated components provide the necessary delay which may be on the order of 1 second. When transistor 124 is saturated, the junction terminal of capacitor 137, resistor 138, and Zener diode 139 will be almost at the control voltage of 24 volts during the time of an exposure.

Under normal conditions when the anode current is within the predetermined range of say 8 to 12 milliamperes per the aforementioned example, the transistor 122 operating in a current amplifier will have a collector voltage just over 10 volts and the voltage divider 140 will provide a bias to a transistor 142 such that it will not be conducting current. The voltage will be divided by resistors 144, a variable resistor 146, another resistor 148, and a Zener diode 150. The Zener diode has a predetermined breakdown value, for example 12 volts,

which must be exceeded before current will flow therethrough. Under the aforementioned normal conditions of operation of the transistor 122 the Zener diode 150 will not permit the conduction of current. Therefore, bidirectional switches 152 and 154 will be in their off or nonconducting condition.

If, however, the X-ray tube beam current rises above the selected predetermined range, the current amplifier transistor 122 will conduct more current causing the conduction of transistor 142 which will in turn energize bidirectional switch 152. The bidirectional switch 152 will short the clockwise winding 110 of the drive motor 106 and also provide a signal to energize the bidirectional switch 156 thereby energizing the main winding 108 to cause clockwise rotation of the drive motor 106. The voltage selected at the variable autotransformer 104 will be increased thereby raising the anode voltage of the X-ray tube 20 causing the output brightness of the phosphor 28 of the intensifier 26 to increase. The increased brightness will in turn cause the output of the operational amplifier 42 to increase in a positive direction thereby decreasing the output of the filament regulator 60 and hence decrease the beam current of the X-ray tube 20.

The circuitry for controlling accelerating voltage detects an increase in the preset X-ray beam current and when such current exceeds a predetermined range the drive motor 106 is caused to position the autotransformer 104 in such a direction as to increase the accelerating voltage to the X-ray tube 20. The resulting increased brightness will then in turn be detected by the photomultiplier 32 and through the action of the operational amplifier 42 and filament regulator 60 the beam current of the X-ray tube 20 is diminished.

Similarly, should the beam current of the X-ray tube fall below the predetermined range, the collector current of transistor 122 will decrease, transistor 142 will be turned off, and a further decrease of beam current will cause the transistor 122 to be less conductive until the voltage across the Zener diode 150 is sufficient to break down the diode and allow current therethrough which renders the bidirectional switch 154 conductive. The bidirectional switch 154 shorts the counterclockwise winding 122 and turns on the bidirectional switch 156 to cause the drive motor 106 to operate in a counterclockwise direction to diminish the accelerating voltage. A diminishing accelerating voltage will in turn require a higher beam current as sensed by the photomultiplier 34 to allow the brightness of the image to remain stabilized.

The circuitry for controlling accelerating voltage as a function of anode current has been described. As the anode current monitored by the metering bridge 120 goes outside a predetermined range, the accelerating voltage will be controlled to bring the anode current back within the desired range. It will be recalled however that as the accelerating voltage is increased the noise level also increases to an extent that may be detrimental to optimum imaging. Therefore, further in accordance with the present invention means are provided for translating the predetermined range of anode current upward as the accelerating voltage is increased. By allowing the anode current range to translate upward, the required accelerating voltage will not be as great. Referring again to FIG. 2, resistors 124 and variable resistor 126 are connected across the emitter resistor 123 of the current amplifier 122. As the setting of the autotransformer 104 is varied by the drive motor 106 the variable resistor 126 will also vary in its magnitude. The variable resistor 126 is coupled to the drive motor 106 in the same manner that couples the autotransformer 104. Thus, as the accelerating voltage is increased the magnitude of variable resistor 126 is also increased thereby increasing the emitter degeneration of transistor 122. The result is that the range of monitored anode current is translated upward so that a larger magnitude of anode current will be required to further increase the accelerating voltage. The translated range of beam current will therefore result in stabilization of the anode current by the filament regulator 60 at a greater magnitude.

Started another way, the more resistance in the emitter circuit of the transistor 122 the less gain provided thereby and hence a greater magnitude of monitored anode current will be necessary to cause transistor 122 to conduct. The lower limit of the translated range will result in the transistor 122 becoming less conductive at a larger lower extreme with the result that the Zener diode 150 will become conductive causing an accordingly decreased accelerating voltage.

Changing the position of switch 44A through E will alter the servo-sensitivities, operating points, and brightness requirements of the stabilizer. In the illustrative embodiment of FIG. 2 it is assumed that the operating mode of 120 frames per second will demand anode current beyond the phase reversal point of the X-ray generator. Hence, when the mode selector switch is positioned to the opposite extreme illustrated, namely to 120 frames per second, the filament current will be preset by the choice of resistor 72 associated with that portion of the mode selector switch identified at 44B. The filament regulator 60 will have no effect on the system except to stabilize the filament current of the X-ray tube 20 to the preset value. Then, as the brightness requirements are changed as a function of the size of the patient or target, the output of the amplifier integrator 42 will be fed directly to the base of transistor 122 through a single stage inverter 121 by means of those portions of the mode selector switch identified at 44C and 44D. The stabilizer will operate only on the basis of controlling the accelerating voltage. That is, as brightness requirements change, only the autotransformer 104 will be altered. The anode current will not be changed since for this mode of operation it has been preset to preclude sense reversal from occurring.

While the present invention has been described with a degree of particularity for the purpose of illustration, it is to be understood that all modifications, substitutions, and modifications within the spirit and scope of the present invention are herein meant to be included. For example, while transistors, Zener diodes, and bidirectional semiconductor switches have been shown for the purpose of illustration, it is to be understood that any suitable devices exhibiting similar characteristics may be utilized. The autotransformer 104 has been illustrated to be utilized for both fluoroscopic and cine modes of operation. When desired, separate transformers may be employed, a small unit for fluoroscopic operation only and a very large unit for cine modes of operation. Additional switches and alternate motor and transformer drives may also be used when desired even though the simplified version utilizing a single autotransformer has been illustrated.

We claim as our invention:

1. A brightness stabilizer with improved image quality comprising, in combination; an X-ray tube; image-producing means; stabilizing means responsive to the brightness of the image for varying the magnitude of the beam current of the X-ray tube; and means responsive to the magnitude of beam current when outside a predetermined range for controlling the accelerating voltage of the X-ray tube.

2. The combination of claim 1 further comprising means for translating said predetermined range upward as the accelerating voltage is increased.

3. The combination of claim 2 wherein the controlling means includes a motor-driven autotransformer and the translating means includes means gang operated with said autotransformer for decreasing the gain of said controlling means to thereby increase the magnitude of the extremes of anode current for translating said predetermined range of anode current.

4. The combination of claim 1 wherein said last-mentioned means includes first means for increasing the accelerating voltage when the beam current exceeds said predetermined range and second means for decreasing the accelerating voltage when the beam current is less than said predetermined range.

5. The combination of claim 4 wherein said last-mentioned means further includes means for monitoring the beam current.

6. The combination of claim 5 wherein said means for monitoring includes voltage divider means for selectively rendering said first means and said second means operative and current amplifier means responsive to the monitored anode current for controlling voltage across said voltage divider means.

7. The combination of claim 6 wherein said current amplifier means actuates said voltage divider means only when the monitored anode current is outside said predetermined range.

8. The combination of claim 5 wherein said means for monitoring the beam current includes a diode bridge.

9. The combination of claim 4 wherein said first means includes a semiconductor switch and said second means includes a Zener diode.

10. The combination of claim 4 wherein said last-mentioned means further includes a motor-controlled autotransformer and switching means for energizing said motor-controlled au-

totransformer to increase the accelerating voltage when said first means is conductive and to decrease the accelerating voltage when said second means is conductive.

11. The combination of claim 1 further comprising operating mode selection means for setting brightness levels and magnitude ranges of filament current, anode current, and accelerating voltage to said X-ray tube for fluoroscopic and cinefluoroscopic operation.

12. The combination of claim 1 including means for presetting said predetermined range for different modes of fluoroscopic operation.

13. The combination of claim 1 including a time delay circuit for enabling said last-mentioned means after sufficient time has elapsed to allow said stabilizer means to reach steady state conditions.

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